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ENCODATA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
ENCODCD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	4
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Search History

DATE: Tuesday, April 22, 2003 Printable Copy Create Case

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L22	L21 and (encod$4)	10	L22
L21	L20 and (phas$4)	22	L21
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L7	L6 and ((two or both) with (first or second or another or plurality or group or "multi" or multiple or additional or set))	203	L7
L6	L5 and ((first or second or another or plurality or group or "multi" or multiple or additional or set) with ((RF or (radio adj frequency) or radio-frequency or radiofrequency or transmission or transmit$5 or excit$7) with coil))	237	L6
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L1	((magnetic adj resonance) or MRI or NMR)	146053	L1

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☐ 1. Document ID: US 20020125888 A1

L11: Entry 1 of 2 File: PGPB Sep 12, 2002

PGPUB-DOCUMENT-NUMBER: 20020125888
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020125888 A1

TITLE: Magnetic resonance imaging apparatus

PUBLICATION-DATE: September 12, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Visser, Frederik	Eindhoven		NL	
Haans, Paulus Cornelius Hendrikus Adrianus	Eindhoven		NL	
Van Den Brink, Johan Samuel	Eindhoven		NL	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw. Desc | Image

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☐ 2. Document ID: US 6501274 B1

L11: Entry 2 of 2 File: USPT Dec 31, 2002

US-PAT-NO: 6501274
DOCUMENT-IDENTIFIER: US 6501274 B1

TITLE: Magnetic resonance imaging system using coils having paraxially distributed transmission line elements with outer and inner conductors

DATE-ISSUED: December 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ledden; Patrick	Malden	MA		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw. Desc | Image

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Term	Documents
HEAD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1352647
HEADS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	229336
(10 AND HEAD).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	2
(L10 AND (HEAD)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	2

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L11: Entry 2 of 2 File: USPT Dec 31, 2002

DOCUMENT-IDENTIFIER: US 6501274 B1
TITLE: Magnetic resonance imaging system using coils having paraxially distributed transmission line elements with outer and inner conductors

Abstract Text (1):
A magnetic resonance imaging system comprises: a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis. The housing contains: a transmit/receive inductor system having a coil about the axis in proximity with the housing, a gradient inductor system having a coil operatively associated with the transmit/receive inductor system, and a field inductor system having a coil operatively associated with the transmit/receive inductor system. The field coil establishes a supervening field about the entire system. The gradient coil initiates perturbations in the fields and produces signals derived responsively from the perturbations. The transmit/receive coil includes a series of electrical transmission line elements paraxially distributed with respect to the axis about the subject. Each transmission line element includes an outer conductor and an inner conductor spaced radially from the outer conductor relative to the axis. The transmit/receive coil initially transmits to the subject a radio frequency energy field and responsively receives from the subject a magnetic resonance energy field. The signals produced correspond to spatial indicia derived from the subject and are presented as such by a master controller.

Parent Case Text (2):
The applicant herein claims the benefit of U.S. Provisional Patent Application No. 60/159,662, dated Oct. 15, 1999 for HIGH RESOLUTION MAGNETIC RESONANCE IMAGING SYSTEM in the name of Patrick Ledden, the applicant herein.

Brief Summary Text (3):
The present invention relates to magnetic resonance imaging and, more particularly, to a high resolution magnetic resonance imaging system and to components thereof.

Brief Summary Text (5):
Magnetic Resonance Imaging (MRI) has proven to be an enormously useful technology both for the detection and diagnosis of human disease as well as for research into the understanding of basic animal physiology. However, current MRI equipment has been limited by achievable signal-to-noise ratio (SNR) and by limitations in the ability to generate homogenous transmit fields for signal excitation, particularly at high magnetic field strengths.

Brief Summary Text (6):
For the acquisition of data from a nuclear magnetic resonance (NMR) signal, four separate components are required. First a static magnetic field must be generated by a permanent magnet generally of the superconducting type. Pursuant to quantum mechanics, the presence of the static magnetic field causes in a subject an energy difference between atomic spins aligned with and against this static magnetic field. The magnitude of the energy difference depends on a variety of factors, including strength of the magnetic field, size of the magnetic moments of individual atomic nuclei, and temperature. In general, a majority of the atomic spins will align with the static magnetic field and a higher energy minority of the atomic spins will align against it. When exposed to an oscillating magnetic field of proper frequency, such as is generated by an alternating current in a radio frequency (RF) coil, some of the lower energy spins aligned with the static magnetic field will be excited to the higher energy state of being aligned against the field. Once the applied transmit RF magnetic field is removed, these excited spins will decay to the lower

energy state of alignment with the static magnetic field. During the decay, these spins will generate their own RF magnetic field, which can be electronically detected by the same or a different RF coil and thereby be characterized. In order to determine spatial information about the quantity and properties of the atomic nuclei of the subject, a second set of coils, gradient coils, are used to perturb the static magnetic field. By generating magnetic field gradients, current in this separate set of coils spatially changes the oscillation frequency of the atomic spins by changing the frequency of the nuclear magnetic resonance (NMR) oscillation at appropriate times during transmit and receive, and spatial information regarding the atomic spins can be decoded and converted into an image. The generation and reception of the NMR signal in the RF coil and the currents in the gradient coils are controlled by a computer system which processes the information obtained and displays it on a computer screen or printed film for human interpretation.

Brief Summary Text (7):
The advantages of using NMR are several-fold. First, information can be obtained non-invasively on a wide variety of in vitro and in vivo subjects. The lack of non-ionizing radiation is particularly attractive when images are obtained from human subjects. Second, the properties of the magnetic spins are extremely sensitive to their surrounding chemical environment. This allows a great deal of information to be determined from the magnetic resonance signal, including chemical and molecular structure of a wide variety of materials as well as the chemical and structural characteristics of animal and human tissue. By obtaining spatially dependent information regarding the NMR signal, it is possible to obtain detailed images, which not only show great anatomic detail, but which also depend on the chemical properties of tissue. This provides additional image contrast, allows improved discrimination between healthy and diseased tissue, and permits researchers to obtain previously unavailable information regarding in vivo physiologic function.

Brief Summary Text (8):
Despite the multiple advantages of MRI, one major limiting factor in the usefulness of the NMR machine is the small magnitude of the NMR signal generated by a subject's nuclei themselves. This weak signal is easily obscured by the noise present in all electronic detection devices. The presence of this noise then limits the maximum achievable resolution or sensitivity of the NMR machine, specifically, its ability to resolve small anatomic details or to characterize time dependent changes in signal intensity, which are important for understanding of a subject's physiology.

Brief Summary Text (9):
In principle, one can improve the sensitivity of the NMR device by increasing the strength of the static magnetic field. While this does increase signal to noise ratio (SNR), it adds problems in terms of the interaction of high frequency magnetic fields and human tissue, leading to difficulties in achieving uniform image quality and even excitation of NMR spins. Simply increasing the magnetic field strength is a very expensive option: a 3T (3-Tesla) human size magnet costs roughly five times that of a 1.5T magnet. In general, such increased cost places a premium on maximizing SNR at a given field strength.

Brief Summary Text (10):
Most of the noise in human MRI comes from the resistance associated with conductive tissue within the human body. As this resistance is roughly proportional to volume of tissue, large coils, which couple to larger volume of tissue, inherently produce lower quality images than smaller coils. While sensitivity can be improved by making smaller coils, there is a limit to this approach in that eventually the desired body part or region of interest will not fit within the coil or field of view of the coil.

Brief Summary Text (11):
One prior art method designed to increase the field of view of small coils is to use multiple coils arranged in a "phased array" (U.S. Pat. No. 4,887,039). In this method, the images from each individual coil are processed separately and then combined in such a fashion as to maximize image quality. While this is a useful strategy, it has certain limitations. First the individual coils need to be carefully oriented to minimize their respective coupling. Despite proper orientation, there always will be residual coupling between four or more coils limiting the maximum number of coils and consequently the gains in sensitivity. Furthermore, in the standard geometry feasible with surface coils, this arrangement still produces inhomogeneous images, which can complicate their interpretation for

diagnostic purposes.

Brief Summary Text (12):
A second problem is the efficient and uniform excitation of the NMR spins. For most imaging sequences, a homogenous excitation of all spins is required. In general, this requires a larger coil, which then reduces the sensitivity of the system. One commonly used technique is to use a larger coil, optimized for transmit with a second coil specialized for receive. However such systems, as presently implemented, suffer from several disadvantages, particularly when used in high field systems. One disadvantage of current volume transmit coils is the inability to control the field to compensate for variations in patient size and position. While, in principle these variations can be accomplished by manually tuning the coil (see J. Thomas Vaughan, Hoby P. Hetherington, Joe O. Out, Jullie W. Pan, Gerald M. Pohost, "High Frequency Volume Coils for Clinical NMR Imaging and Spectroscopy", Magnetic Resonance in Medicine 32:206-218 (1994)) or by using electromechanical relays to switch in additional reactive circuit elements, such methods are time consuming and subject to the variability of mechanical connections.

Brief Summary Text (13):
Conventional MRI coils come in two basic categories. (1) The simpler, the surface coil, consists of one or more conductive loops. Additional reactive circuit components, such as capacitors and inductors, are used to tune the coil and couple energy to or from it to the rest of the NMR system. Importantly, active circuit elements, such as PIN diodes, can be added to allow specialization of coil function for receive or transmit. (2) Volume coils, such as birdcage coils, consist of one or more large surface coils oriented in such a fashion as to produce a homogenous magnetic field. While such coils are in common use, the large size of these coils makes them poor receivers of NMR signal. This difficulty can be overcome by using PIN diodes to "detune" the volume coil for use with a more sensitive surface coil receiver.

Brief Summary Text (14):
In particular, at high fields, the use of volume coils becomes increasing problematic. The large size of these coils required to enclose a useful area of human anatomy, such as the torso or head, leads to them becoming efficient radiators of electromagnetic energy. Moreover, the interaction of large volume coils with tissue at high frequencies leads to non-uniform magnetic fields within human tissue complicating the ability to obtain uniform spin excitation.

Brief Summary Text (15):
The following U.S. Pat. No. 5,557,247 to Vaughn, U.S. Pat. No. 4,751,464 to Bridges, U.S. Pat. No. 4,746,866 to Roschmann and U.S. Pat. No. 4,506,224 to Krause, disclose volume coils based on cavity resonators. Conductive segments within the cavity interact to form a resonant structure. While this coil can offer improved efficiency over a conventional volume coil, several disadvantages exist. First, the structure being closed can give a subject a sense of claustrophobia and make it difficult to present visual stimulation for research purposes. Second, the closed shielded nature of the coil makes it difficult to specialize for the use of transmit or receive purposes. If circuit elements are added to detune the coil, the outer cavity shield will interact with smaller coils placed with the larger cavity, impairing their performance. Additionally, the cavity shield prevents the use of the coil for specialization as a smaller coil to use with receive only function or as its use as a phased array.

Brief Summary Text (17):
The present invention is an improved NMR coil design based on the use of transmission line segments rather than conventional inductive coil elements. The use of transmission lines has several benefits. Transmission lines have a concentration of electromagnetic fields between their elements. By adjusting the distance between these conductive elements, interaction of the magnetic fields of the transmission line with an external sample can be controlled and optimized for NMR signal generation and/or detection. The presence of two conductors also decreases the inductance of each conductor. This minimizes the electric fields associated with the conductors, which is advantageous since these electric fields can be associated with dielectric tissue losses which decrease coil efficiency and sensitivity. Moreover, the inherent shielded nature of transmission lines decreases the radiation of electromagnetic energy from the NMR coil, improving coil efficiency and sensitivity over conventional NMR coil design. The shielded nature of a transmission line also decreases the interaction or coupling between coil elements. This can be

advantageous since under proper conditions, coil elements can operate with minimal interaction. This allows a large single coil structure to operate as multiple smaller individual coils. With proper combinations, these separate coils can be combined in such a way to optimize NMR signal generation and/or reception. In particular, by combining signals from individual coil elements, spatial information may be decoded regarding the NMR signal, increasing the sensitivity and speed of data acquisition for both high field and low field NMR systems.

Brief Summary Text (18):
The coil consists of N transmission line segments distributed in a circular, elliptical, or other geometrical arrangement. Each transmission line element is comprised of two or more individual conductors with or without additional lumped or distributed capacitive or inductive circuit components. In general, each transmission line element couples to the others through mutual inductance and capacitive coupling. Additional lumped or distributed inductive or capacitive elements may be placed between the transmission line segments to alter this coupling. The combined influences of the interaction between these elements gives rise to frequency dependent relations between the currents and voltages present on individual transmission line elements. By changing the individual circuit components and transmission line geometry, a given current distribution can be obtained on the transmission line elements at a given frequency. The magnetic field arising from the currents on each element add through superposition and create a given magnetic field configuration for use in either or both the generation and detection of the NMR signal.

Brief Summary Text (19):
In particular, with placement of properly valued reactive components between individual transmission line elements, mutual coupling between elements can be minimized. This allows the resonant structure of the N transmission line segment to become degenerate and allows the currents on each element to be relatively independent. This has the advantage for NMR signal generation in that the currents on each element can be individually controlled at will to generate a excitation field of a desired spatial and phase characteristic. Additionally in such an degenerate mode arrangement, received signals from each element are independent and can then be combined in such a way to optimize image homogeneity, sensitivity, or other desired parameters.

Brief Summary Text (20):
In order for the transmission line structure to be useful, energy needs to be transferred into the coil during signal generation and out of the coil during signal reception. This can be accomplished by inductively or capacitively coupling one or more circuit elements to one or more RF power amplifiers and/or RF receivers. This coupling can be adjusted to allow an arbitrary impedance of such equipment to be matched to the currents and voltages found in the transmission line structure. In particular, the phases of the current in two or more transmission line elements can be offset as to create elliptically polarized magnetic fields for improved efficiency in the generation and/or detection of the nuclear magnetic resonance signal.

Brief Summary Text (21):
In addition to passive components, active circuit elements such as diodes (either regular or PIN) can be added to this structure. With diodes, the tuning of individual transmission line elements or their mutual coupling can be changed in order to modify the current distribution and element impedance of the transmission line segments. When used with one or more additional coils (which may be a combination of transmission line structures or conventional NMR coils), these diodes can be arranged so that during transmit or receive functions, one coil has a desired magnetic field configuration while the other coil presents a high impedance so as not to interfere with the magnetic fields of the first coil. In this manner, each of the two or more coils can be optimized for either transmit or receive, resulting in improved generation and detection of the NMR signal.

Brief Summary Text (22):
Other active circuit elements can be added to the transmission line structure such as vacuum tubes or transistors (including but not limited to conventional bipolar transistors, field effect transistors, gallium arsenide field effect transistors, high electron mobility transistors, pseudomorphic high electron mobility transistors, or heterojunction bipolar transistors). These transistors can be used to provide amplification of either the transmit energy needed in the generation of

the nuclear magnetic signal or the small magnitude received energy from the NMR spin decay. In this way, signal losses arising from matching circuits and connecting cables are minimized, leading to improved coil efficiency. If the coil is designed for both transmit and receive functions, diodes may be included to change the coupling between these active amplifier circuits and individual transmission line elements. In this manner, transistors designed for low-noise signal amplification are not damaged by the high element currents during the transmit function and transistors circuits designed for power amplification do not add noise during signal reception.

Brief Summary Text (23):
The addition of active vacuum tube or transistor circuits can provide additional advantages. With proper design, these circuits can present impedance mismatches to the transmission line structure while simultaneously preserving adequate amplifier function. These impedance mismatches can be used to change or minimize coupling between individual transmission line elements, allowing the elements to be decoupled and be relatively independent of each other. During transmit, this has the advantage that individual element currents can be changed electronically in magnitude or phase so as to modify the desired magnetic field for optimal transmit excitation without requiring change or variation of passive circuit elements. This is particularly advantageous at high frequencies where dielectric resonances in human tissue require non-uniform magnetic fields for uniform spin excitation. Additionally, during receive, decoupling of the currents on transmission line elements allows each element to function as a separate signal detector. By combining the signals from these elements electronically, either directly after amplification or at a later stage such as after image reconstruction, these signals can be added in a way such that sensitivity is maximized for one or more areas of interest. In particular, the spatially dependent information from each element can be combined after image reconstruction in such a manner that sensitivity is maximized at each point in an image.

Brief Summary Text (24):
Moreover, the geometric arrangement of the individual transmission line elements can be used to decode spatial information regarding the detected NMR signal. By decoding spatial information from individual coil elements, the steps required for the acquisition of an NMR image can be reduced, allowing the imaging process to be completed in less time.

Brief Summary Text (25):
The illustrated embodiments of the present invention demonstrate an actively decoupled transmission line resonator for use as a transmit coil in conjunction with surface coil receivers, as well as use of a transmission line structure as a receive array coil.

Drawing Description Text (4):
FIG. 1a illustrates a partial assembly of a transmission line coil, with one line element and the rear end plate;

Drawing Description Text (7):
FIG. 4 illustrates a completed coil incorporating the transmission line assembly of FIG. 2 and the frame of FIG. 3;

Drawing Description Text (8):
FIG. 5 is an end view of a detunable head transmit coil embodying the present invention;

Drawing Description Text (9):
FIG. 6 is a side view of the detunable head transmit coil of FIG. 5;

Drawing Description Text (14):
FIG. 11 illustrates an image produced by a head transmit coil for both transmit and receive in accordance with the present invention;

Drawing Description Text (15):
FIG. 12 illustrates an image produced by a head transmit coil for transmit and a dual loop coil for receive in accordance with the present invention;

Drawing Description Text (16):
FIG. 13 illustrates a human brain image produced by a head transmit coil used for

both transmit and receive;

Drawing Description Text (17):
FIG. 14 illustrates a human brain image produced by a head transmit coil used for transmit and a loop coil for receive;

Drawing Description Text (19):
FIG. 18 illustrates a power amplifier directly connected to an individual transmission line element in accordance with the present invention;

Drawing Description Text (20):
FIG. 19 illustrates a low-noise transistor amplifier directly connected to an individual transmission line element in accordance with the present invention;

Drawing Description Text (21):
FIG. 20 illustrates a combination of active transmit and receive circuits directly connected to an individual transmission line element in accordance with the present invention;

Drawing Description Text (22):
FIG. 21 illustrates an overall system diagram in connection with a transmission line coil used for both transmit and receive in accordance with the present invention;

Drawing Description Text (23):
FIG. 22 illustrates an overall system diagram in connection with separate transmit and receive coils in accordance with the present invention;

Drawing Description Text (24):
FIG. 23 illustrates an overall system diagram in connection with a transmission line coil used in array mode in accordance with the present invention;

Drawing Description Text (25):
FIG. 24 illustrates images obtained from four element transmission line coil operated in the receive array mode whereby low noise preamplifiers detune the interactions between elements allowing each transmission line element to function independently;

Drawing Description Text (27):
FIG. 26 illustrates human images obtained from an elliptic four element transmission line coil operated in the receive array mode whereby a combination of capacitors and low noise preamplifiers detune the interactions between elements allowing each transmission line element to function independently; and

Detailed Description Text (3):
Generally, the illustrated system of the present invention includes a housing 11 within which are a field coil 13, X, Y, Z gradient coils 15 and a transmit/receive coil 17. Field coil 13 is energized by a field coil controller 19. Gradient coils 15 are controlled by a gradient coil controller 21. Transmit/receive coil 17 is controlled by a transmit/receive coil controller 23. Field coil controller 19, gradient coil controller 21, and transmit/receive controller 23 are managed by a system controller 25. A carriage 27, upon which a subject reclines, is reciprocable into and out of the region 29 within transmit/receive coil 17. Region 29 is a medical diagnostic chamber within which the subject is internally imaged pursuant to the present invention.

Detailed Description Text (4):
As shown in FIGS. 1a to 4, the coil at the heart of the present invention, is a transmit and/or receive coil that includes transmission line elements distributed in a circular, elliptical, or other geometrical arrangement. FIG. 1a, for simplicity, shows a single transmission line element 12a (collectively, 12) mounted to a circular end cap 14 with slots 22. The present invention contemplates that the end cap 14 may have another shape, such as a dome, or may be absent altogether. FIG. 2 shows a complete transmission line assembly 16 with all 16 transmission line elements 12. Note that there are gaps 24, corresponding the end cap slots 22, between each of the elements 12. FIG. 3 shows a mechanical frame 18 into which the transmission line assembly 16 is fixed. FIG. 4 shows an entire coil structure 10 including an outside case or housing 20.

Detailed Description Text (5):

Each transmission line element 12 comprises two or more individual conductors 26, 28 with or without additional lumped or distributed capacitive or inductive circuit components. Each transmission line element 12 couples to the others through mutual inductance and capacitive coupling. Additional lumped or distributed inductive or capacitive elements, in various embodiments, are placed between transmission line segments (groups of transmission line elements 12) to alter this coupling. The combined influences of the interaction between the elements 12 give rise to frequency-dependent relations between the currents and voltages present on individual transmission line elements 12, as shown in the Kirchoff circuit relationship between transmission line currents and voltages: ##EQU1##

Detailed Description Text (6):
where Xelem.sub.j is the complex impedance of transmission line element n at frequency .omega., Xm.sub.i,j is the complex impedance associated with the coupling between elements i and j, I.sub.n is the current through element n, and V.sub.n is the voltage across element n. By changing the individual circuit components and transmission line geometry, a given current distribution can be obtained on the transmission line elements 12 at a given frequency. The magnetic field arising from the currents on each element 12 add through superposition to create a given magnetic field configuration for use in either or both the generation and detection of an NMR signal.

Detailed Description Text (7):
In order for the transmission line structure to be useful, energy must be transferred into the coil 10 during signal generation and out of the coil 10 during signal reception. This can be accomplished by inductively or capacitively coupling one or more elements 12 to one or more RF power amplifiers and/or RF receivers. The coupling can be adjusted to allow an arbitrary impedance of such equipment to be matched to the currents and voltages found in the coil 10. In particular, the phases of the current in two or more transmission line elements 12 can be offset so as to create circularly or elliptically polarized magnetic fields for improved efficiency in the generation and/or detection of the NMR signal.

Detailed Description Text (8):
In addition to passive components, in various embodiments, active circuit elements such as diodes (either regular or PIN) are added to the circuit. With diodes, the tuning of individual transmission line elements 12 or their mutual coupling can be changed in order to modify the current distribution and element impedance of the transmission line elements 12. When used with one or more additional coils 10 (which may be a combination of transmission line structures or conventional NMR coils), these diodes can be arranged so that, during transmit or receive functions, one coil 10 has a desired magnetic field configuration while another coil 10 presents a high impedance, so as not to interfere with the magnetic fields of the first coil 10. In this manner, each of the two or more coils 10 can be optimized for either transmit or receive, resulting in improved generation and detection of the NMR signal.

Detailed Description Text (9):
In addition to diodes, in various embodiments, other active circuit elements are added to the transmission line structure 16. These include vacuum tubes or transistors (including but not limited to conventional bipolar transistors, field effect transistors, gallium arsenide field effect transistors, high electron mobility transistors, or heterojunction bipolar transistors). Transistors can be used to provide amplification of either the transmit energy needed in the generation of the NMR signal or the small received energy from the NMR spin decay. In this way, signal losses arising from matching circuits and connecting cables are minimized, leading to improved coil efficiency. If the coil 10 is designed for both transmit and receive functions, diodes may be included to change the coupling between the active amplifier circuits and individual transmission line elements 12. In this manner, transistors designed for low-noise signal amplification are not damaged by the high element currents during the transmit function, and transistor circuits designed for power amplification do not add noise during signal reception.

Detailed Description Text (10):
The addition of active vacuum tube or transistor circuits can provide additional advantages. Pursuant to the present invention, these circuits can present impedance mismatches to the transmission line structure 16 while simultaneously preserving adequate amplifier function. Impedance mismatches can be used to change or minimize coupling between individual transmission line elements 12 allowing the elements' currents to be decoupled and relatively independent of each other. During transmit,

this has the advantage that individual element currents can be changed electronically in magnitude or phase so as to modify the desired magnetic field for optimal transmit excitation without requiring change or variation of passive circuit elements. This is particularly advantageous at high frequencies where dielectric resonance in human tissue require non-uniform magnetic fields for uniform spin excitation.

Detailed Description Text (11):
Additionally, during receive, decoupling of the currents on transmission line elements 12 allows each element 12 to function as a separate signal detector. By combining the signals from these elements 12 electronically, either directly after amplification or at a later stage such as after image reconstruction, these signals can be added in such a way that sensitivity is maximized for one or more areas of interest. In particular, the spatially dependent information from each element 12 can be combined after image reconstruction in such a manner that sensitivity is maximized at each point of an image. Moreover, the geometric arrangement of the individual transmission line elements 12 can be used to decode spatial information regarding the detected NMR signal. By decoding spatial information from individual coil elements 12, the steps required for the acquisition of an NMR image can be reduced, allowing the imaging process to be completed more quickly.

Detailed Description Text (13):
FIGS. 5 and 6 show the geometry of a detunable volume transmit coil 40 constructed in accordance with the present invention. The volume coil 40 is a flat element, shielded transmission line design that is analogous to a previously described design utilizing coaxial elements. (See U.S. Pat. No. 4,746,866 to Roschman and U.S. Pat. No. 5,557,247 to Vaughan.) The coil 40 incorporates an end-capped structure that decreases radiation losses. A conductive cavity wall 42 is divided into 12 outer conductors 44 by regularly spaced longitudinal slots 46. The slots 46 minimize eddy currents when used in echo planar imaging (EPI). In one configuration, the diameter of the coil 40 is 37.5 cm and the axial length is 20 cm. Twelve flat copper inner conductors 50, each with a width of 2.5 cm, are located 1.75 cm inwardly from the cavity wall 42, as at 52, and centered between the slots 46. The inner conductors 50 are tuned using nonmagnetic chip capacitors and nonmagnetic tuning capacitors. Two elements 12a, 12d, located 90.degree. from each other, are matched to 50.OMEGA. using lumped element quarter-wave transformers. These outputs then are driven through a quadrature coupler.

Detailed Description Text (14):
In one configuration, shown in FIG. 7, detuning is accomplished with a diode 60. The circuit shown in FIG. 8 utilizes a shunt diode configuration similar to that described in Ledden, P. J., Wald, L. L., Vaughan, J. T., "Volume Coil Transmit Surface Coil Receive System for Brain Imaging at 3T", Proceedings of the International Society of Magnetic Resonance in Medicine, p. 168 (1999). In this arrangement, a diode 62 is placed across the tuning capacitor 64 at the posterior end of every element 12. During transmit, the diodes 62 are back biased and have a high impedance allowing normal tuned coil operation. During receive, the diodes 62 are forward biased, shorting the tuning capacitors 64 and detuning the coil 10. Bias voltage is applied to each diode 62 through high impedance RF chokes 66 which have an RF impedance of greater than 1 K.OMEGA. at 127.8 MHz. The diode bias voltage is provided by a coaxial cable 68 separate from the RF connections and is controlled by a 5 V digital signal from a scanner.

Detailed Description Text (15):
In another configuration, shown in FIG. 9, detuning utilizes a lumped element quarter wave line 76 between the diode 72 and the tuning capacitor 74. In this arrangement, the diode 72 is forward biased during transmit and shorts the quarter wave circuit 76. This causes the quarter wave circuit 76 to present a high impedance across the tuning capacitor 74 allowing normal coil resonance. During receive, the diode 72 is back biased and causes the quarter wave circuit 76 to short the tuning capacitor 74, thereby detuning the coil 10. Unlike the configuration in FIG. 9, high negative bias voltages are not required, since during receive, the RF voltages in the coil 10 are very small. This eliminates the need for a high voltage bias supply and driver resulting in improved operator and patient safety.

Detailed Description Text (16):
Two different geometries of receive-only surface coils are presented herein. The first coil consists of a quadrature surface coil 70 comprising two 9-cm loops 72, as in FIG. 10. Of similar overall design, the second coil consists of two 12

cm.times.20 cm curved rectangular loops, also combined in quadrature. Each coil is matched to 50.OMEGA. using a standard balun drive circuit 74. Detuning during transmit is accomplished by placing a PIN diode 76 across the balun 74. In the conductive state, this diode 76 shorts the balun 74, causing the coil 70 to double tune with a null at the 128 MHz.

Detailed Description Text (17):
All electrical impedance measurements in the aforementioned embodiments were made with a network analyzer. The isolation produced by the diode detuning of the transmit coil was determined by the change in radio frequency transmission between two untuned 2.5-cm-diameter probe coils loosely coupled to the volume coil. The isolation was taken as the difference in radio frequency insertion loss in decibels at 127.8 MHz between the tuned and detuned states. The two probe coils were physically separated and made electrically orthogonal to minimize their inductive coupling. A similar method was used to measure the degree of detuning obtained by the active PIN diode trap structure on the receive surface coils.

Detailed Description Text (19):
All studies were performed using a 3T system incorporating an 80-cm bore magnet, and a resonant gradient coil for EPI. Coil SNR was calculated by dividing the image intensity by the standard deviation of the background noise. Transmit efficiency was compared to the standard commercial 16-rung birdcage coil (28 cm diameter and 30 cm length) provided with the system by comparing the transmit gain required for 90.degree. spin echo excitation. All human studies were conducted with Institutional Review Board (IRB) approval.

Detailed Description Text (21):
Measurements of the detuning of the transmit coil were as follows. Greater than 40 dB of isolation was achieved between the tuned and detuned states using either of the diode detuning methods. With careful adjustment, the PIN diode trap circuit on the receive coils also provided greater than 40 dB of isolation in the detuned state. Less than 100 kHz change in loaded resonant frequency occurred when either receive-only surface coil was placed within the detuned transmit coil.

Detailed Description Text (23):
The transmit power required for a 90.degree. pulse excitation for the transmission line resonator was compared to a standard commercial birdcage coil. Despite its larger size, the transmission line resonator without detuning circuitry had approximately 10% greater efficiency than did the birdcage design. Some loss of coil efficiency occurred with either detuning circuit. Addition of the direct shunt diode configuration reduced the transmit efficiency by an amount dependent on the reverse bias voltage applied. Conversely the quarter wave diode short configuration required only enough bias voltage to forward bias the diode, but coil efficiency depended on bias current. With the direct shunt diode circuit, coil efficiency was reduced approximately 0.5 db with -250 V diode back bias. In comparison, the quarter wave diode short circuit reduced coil efficiency 75 db with 200 mA forward bias current.

Detailed Description Text (24):
FIGS. 11 and 12 show phantom images taken with the head transmit coil system. FIG. 11 shows the volume coil transmitter being used both for transmit and receive. With the relatively large size of the head transmit coil, highly uniform transmit excitation was achieved. FIG. 12 shows the results when the detunable volume coil was used for transmit and a dual 9-cm loop pair was used for receive. Image intensity decreased smoothly as a function of distance from the receive coil elements, indicating the absence of surface coil focusing of the transmit fields and good detuning of the transmit coil during receive.

Detailed Description Text (25):
FIGS. 13 and 14 show other results obtained with a human subject. In FIG. 13, the transmit coil was used for both transmit and receive. The relatively large size of the transmit coil results in a uniform transmit field over the human brain. Image SNR was approximately 10% greater than with the birdcage head coil. FIG. 14 shows the results obtained when the detunable volume coil was used for transmit and the dual 9-cm loop pair was used for receive. As with the phantom image, the surface coil receivers markedly increased local SNR. In comparison to the birdcage head coil, the combination of the head transmit coil and receive-only 9-cm loop pair provided up to 500% improvement in cortical SNR.

Detailed Description Text (27):

These results demonstrate the feasibility of a volume coil transmit, surface coil receive system for brain imaging at 3T. Despite the high frequency and close proximity to the surface coil, adequate isolation was achieved between the detuned transmit resonator and the surface coil receiver during both transmit and reception. The receive-only surface coil provided a marked increase in local SNR.

Detailed Description Text (30):
The phantom and human images demonstrate the feasibility of PIN diode detuning of high frequency transmission line resonators for use with surface coil receivers at 3T. The use of small local surface coil receivers allow improved SNR for a wide variety of brain imaging applications at 3T and enable full utilization of the increased sensitivity of high field MR systems.

Detailed Description Text (31):
As indicated above, the present invention contemplates the connection of active transistor amplifiers directly to one or more of the transmission line elements. Thus, FIG. 18 shows a power transistor amplifier 80 directly connected to an individual transmission line element 12. FIG. 19 shows a low-noise transistor amplifier 82 directly connected to an individual transmission line element 12. FIG. 20 shows a combination of an active transmit circuit 84 and active receive circuit 86 directly connected to an individual transmission line element 12 with PIN diode switching. Transistor and PIN diode DC connections are omitted for clarity.

Detailed Description Text (33):
FIG. 21 is a diagram of a system 100 that uses a transmission line coil used for both transmit and receive. As shown, one or more coil elements 102 are connected through matching circuits directly or through a signal combiner, such as a quadrature combiner, to a transmit/receive (T/R) switch 104. During NMR signal generation, the T/R switch 104 connects the transmission line coil 102 to an RF generator 108 through a power amplifier 106. During NMR signal detection, the T/R switch 104 connects the coil 102 to a signal receiver 112 through a low-noise preamplifier 110. In conjunction with properly timed magnetic field gradients coils 114, the system controller 116 acquires data and processes it into an image or other useful form.

Detailed Description Text (34):
FIG. 22 is a diagram of a system 120 that uses separate transmit coils 122 and receive coils 124. As shown, an RF generator 126 is connected through an RF power amplifier 128 and T/R switch 130 to one or more transmission line coil elements 122. During transmit, PIN diodes in the coils 122 are biased to allow normal tuned coil operation, while PIN diode circuits detune the receive coil 124. During NMR signal detection, the T/R switch 130 connects the receive coil 124 to a signal receiver 132 through a low-noise preamplifier 134. PIN diodes detune the transmit coil 122, while the receive coils 124 are biased for normal tuned operation. In conjunction with properly timed magnetic field gradient coils 136, the system controller 138 acquires data and processes it into an image or other useful form.

Detailed Description Text (35):
FIG. 23 is a diagram of a system 150 that uses a transmission line coil 152 in array mode. As shown, one or more individual transmission line elements 154 are connected through built-in transistor amplifiers and T/R switches to separate NMR receivers 158 and RF generators 156. During NMR signal generation, each T/R switch connects a transmission line coil element 154 to a generator 156. During NMR signal detection, each T/R switch connects a coil element to a separate signal receiver 158. In conjunction with properly timed magnetic field gradient coils 160, the system controller 162 acquires data and processes it into an image or other useful form. By using separate RF signal generators 156 and receivers 158 for each coil element 154, the signals from individual elements 154 can be optimally controlled and processed for maximum advantage.

Detailed Description Text (36):
FIG. 24 shows results obtained from such array system. In this case, separate preamplifier/receiver channels on a General Electric (GE) 1.5T MRI scanner are connected to each of four separate transmission line elements arranged in a cylindrically symmetric fashion similar to the 16 element array shown in FIGS. 1a -4 the impedance mismatch between the low-noise preamplifiers detunes the mutual inductive coupling between elements and allows the currents on each element to be independent. A PIN diode circuit detunes each transmission line element during transmit to allow the larger coil to generate a highly uniform spin excitation

field. The images from each transmission line element are independent and can be processed separately to create an image of desired spatial sensitivity.

Detailed Description Text (37):
FIG. 25 shows the combination of each separate receive channel combined either to create a homogenous image (right) or a gradient mode image (left). Depending on the frequency of operation, a combination of these two images can be used to correct for image intensity variations caused by dielectric resonances or other high-field image artifacts.

Detailed Description Text (38):
FIG. 26 shows the results obtained with a transmission line array coil with a human volunteer. In this case, the four transmission line elements were arranged in an elliptical fashion to more closely fit the geometry of the human head. A combination of capacitors between elements and the low-noise preamplifiers detuned the coupling between elements allowing each to operate independently. As in FIG. 24, a separate preamplifier/receiver was connected to each transmission line element. A PIN diode detuning circuitry allowed the use of a highly homogenous transmit coil to provide uniform spin excitation. As seen in the images, each transmission line element operates independently providing a high sensitivity image of a portion of the human brain.

Detailed Description Text (39):
FIG. 27 shows a sum-of-squares recombination of the images in FIG. 26. By appropriately combining multiple high-sensitivity images obtained from individual transmission line elements, a homogenous image is obtained which has higher local sensitivity than could be obtained with combining signals from each element with the fixed amplitude and phase relationships as found in FIG. 13.

Detailed Description Text (40):
Thus it has been shown and described an improved NMR coil design based on the use of transmission line elements which satisfies the objects set forth above.

Detailed Description Text (41):
Since certain changes may be made in the present disclosure without departing from the scope of the present invention, it is intended that all matter described in the foregoing specification and shown in the accompanying drawings be interpreted as illustrative and not in a limiting sense.

Other Reference Publication (1):
Ledden et al., "Volume Coil Transmit Surface Coil Receive System for Brain Imaging at 3T", Proceedings of the International Society of Magnetic Resonance in Medicine, p. 168 (1999).

CLAIMS:

1. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis; (b) a transmit/receive inductor system about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said transmit/receive inductor system; (d) a static magnetic field inductor system operatively associated with said transmit/receive inductor system; (e) said transmit/receive inductor system constituting a coil having an outer surface about said axis and including a series of electrical transmission line elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface; (f) said coil initially transmitting to said subject fields of radio frequency energy as a transmit signal, and responsively receiving from said subject fields of magnetic resonance energy as a receive signal; (g) said gradient inductor system initiating perturbations in said fields and producing signals derived responsively from said perturbations; (h) said signals corresponding to spatial indicia derived from said subject.

2. The magnetic resonance imaging system of claim 1 wherein said coil establishes concentrations of electromagnetic fields among said transmission line segments.

3. The magnetic resonance imaging system of claim 2 wherein, by adjusting the distance between said transmission line segments, the interaction of the magnetic

fields of said transmission line segments with an external sample can be controlled and optimized for nuclear magnetic resonance signal generation and/or detection.

4. The magnetic resonance imaging system of claim 1 wherein said plural transmission line segments decrease the inductance of each line segment and minimize the electric fields associated therewith, whereby dielectric tissue losses is said subject are reduced.

5. The magnetic resonance imaging system of claim 1 wherein said plural transmission line segments have inherent shielding, whereby coupling between said transmission line segments is controlled.

6. The magnetic resonance imaging system of claim 1 wherein said plural line segments are combined to optimize NMR signal generation and/or reception.

7. The magnetic resonance imaging system of claim 1 wherein signals form said plural line segments are combined to decode spatial information derived from the NMR signal, thereby to increase the sensitivity and speed of data acquisition.

8. The magnetic resonance imaging system of claim 1 wherein said inductor consists of N transmission line segments arranged in a geometric pattern in which said line segments are substantially equidistant from each other.

9. The magnetic resonance imaging system of claim 1 wherein said geometric pattern is circular or elliptical.

10. The magnetic resonance imaging system of claim 1 wherein said geometric pattern is flat or curved.

11. The magnetic resonance imaging system of claim 1 wherein each of said transmission line segments includes at least two individual conductors together with additional lumped or distributed capacitive or inductive circuit components.

12. The magnetic resonance imaging system of claim 1 wherein each transmission line element couples to the others through mutual inductance and capacitive coupling.

13. The magnetic resonance imaging system of claim 1 wherein distributed impedance elements are connected between certain of said transmission line segments to alter the coupling therebetween.

14. The magnetic resonance imaging system of claim 1 wherein impedance elements are connected between said transmission line segments to establish interactions that establish frequency dependent relations between the currents and voltages present on certain of said transmission line segments.

15. The magnetic resonance imaging system of claim 1 wherein a given current distribution is obtained on said transmission line elements at a given frequency by adjustment of the geometry of said transmission line elements and circuit components connected among said transmission line elements.

16. The magnetic resonance imaging system of claim 1 wherein the fields generated by the currents in said transmission line elements are superposed to create a given magnetic field configuration for use in either or both the generation and detection of the NMR signal.

17. The magnetic resonance imaging system of claim 1 including RF power amplifiers and/or RF receivers coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal and out of said coil during the reception of said receive signal.

18. The magnetic resonance imaging system of claim 1 including at least an RF power amplifier reactively coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal, the impedance of said RF power amplifier and the impedance of said one of said transmission line elements being matched.

19. The magnetic resonance imaging system of claim 1 including at least an RF receiver reactively coupled to at least one of said transmission line elements for transferring energy from said coil during the reception of said receive signal, the

impedance of said RF receiver and the impedance of said one of said transmission line elements being matched.

20. The magnetic resonance imaging system of claim 1 wherein the phases of the current in a plurality of said transmission line segments are offset so as to create an elliptically polarized magnetic field for generating and/or detecting nuclear magnetic resonance signals.

21. The magnetic resonance imaging system of claim 17 including a plurality of diodes operatively connected to a plurality of said transmission line segments for tuning the coupling between said transmission line segments and said RF amplifiers and receivers.

22. The magnetic resonance imaging system of claim 1 including reactive coupling elements between one or more transmission line elements to allow the currents on each transmission line element to be relatively independent.

23. The magnetic resonance imaging system of claim 1 with individual preamplifiers connected to each transmission line element with impedance mismatches designed to allow each transmission line element to operate independently allowing the signals from each transmission line element to be combined either before or after image reconstruction for optimal image reception.

24. The magnetic resonance imaging system of claim 1 with individual preamplifier/receivers connected to each transmission line element with the independent information obtained from individual transmission line elements being used to decode spatial information regarding said subject.

25. The magnetic resonance imaging system of claim 1, with individual power amplifiers connected to each transmission line element with impedance mismatches designed to allow the current of each transmission line element to be independently controlled allowing a transmit field of desired spatial intensity and phase to be generated.

26. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber with a static homogenous magnetic field for a subject therewithin lying along an axis; (b) a plurality of transmit/receive inductor systems about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said transmit/receive inductor systems; (d) a static magnetic field inductor system operatively associated with said transmit/receive inductor systems; (e) at least one of said transmit/receive inductor systems constituting a coil having an outer surface about said axis and including a series of electrical transmission line elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface; (f) each said coil selectively transmitting to said subject fields of radio frequency energy, and selectively receiving from said subject fields of magnetic resonance energy; (g) said gradient inductor system initiating perturbations in said fields and producing signals derived responsively from said perturbations; (h) said signals corresponding to spatial indicia derived from said subject.

27. The magnetic resonance imaging system of claim 26, wherein one of said coils is a conventional loop inductor.

28. The magnetic resonance imaging system of claim 26, wherein one of said coils is a conventional loop inductor which is detuned during transmit function, said transmit function being performed by a transmission line coil which is detuned during receive.

29. The magnetic resonance imaging system of claim 26, wherein one of said coils is a phased array of conventional loop inductors.

30. The magnetic resonance imaging system of claim 26, wherein one of said coils is a phased array of conventional loop inductors which are detuned during transmit function, said transmit function being performed by a transmission line coil which is detuned during receive function.

31. The magnetic resonance imaging system of claim 26, wherein one of said coils is

an array of said transmission line elements each operated independently with individual preamplifiers/receivers.

32. The magnetic resonance imaging system of claim 26, wherein one of said coils is an array of said transmission line elements each operated independently with individual preamplifiers/receivers, said array being detuned during system transmit function.

33. The magnetic resonance imaging system of claim 26, wherein said system includes at least two coils, one of said coils being a transmit coil and the other of said coils being a receive coil.

34. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis; (b) a transmit inductor system about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said transmit inductor system; (d) a static magnetic field inductor system operatively associated with said transmit inductor system; (e) said receive inductor system constituting a coil having an outer surface about said axis and including a series of electrical transmission fine elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface, said coil including a means for detuning said coil to prevent disturbance of the transmit fields generated by a separate transmit inductor system; (f) said coil initially transmitting to said subject fields of radio frequency energy as a transmit signal; (g) said gradient inductor system initiating perturbations in said fields.

35. The magnetic resonance imaging system of claim 34 wherein said coil establishes concentrations of transmit electromagnetic fields among said transmission line elements.

36. The magnetic resonance imaging system of claim 34 wherein, by adjusting the distance between said transmission line elements, the interaction of the magnetic fields of said transmission line elements with an external sample can be controlled and optimized for nuclear magnetic resonance signal generation excitation.

37. The magnetic resonance imaging system of claim 34 wherein said series of transmission line elements decrease the inductance of each line element and minimize the electric fields associated therewith.

38. The magnetic resonance imaging system of claim 34 wherein said series of transmission line elements have inherent shielding.

39. The magnetic resonance imaging system of claim 34 wherein said transmit inductor system consists of N transmission line elements arranged in a geometric pattern in which each of said transmission line elements is substantially equidistant from each adjacent transmission line element.

40. The magnetic resonance imaging system of claim 39 wherein said geometric pattern is circular or elliptical.

41. The magnetic resonance imaging system of claim 39 wherein said geometric pattern is flat or curved.

42. The magnetic resonance imaging system of claim 34 wherein said outer and inner conductors include additional lumped or distributed capacitive or inductive circuit components.

43. The magnetic resonance imaging system of claim 34 wherein each of said transmission line elements couples to the other of said transmission line elements through mutual inductance and capacitive coupling.

44. The magnetic resonance imaging system of claim 34 wherein distributed impedance elements are connected between certain of said transmission line elements to alter the coupling therebetween.

45. The magnetic resonance imaging system of claim 34 wherein impedance elements are connected between said transmission line elements to establish interactions that

establish frequency dependent relations between the currents and voltages present on certain of said transmission line elements.

46. The magnetic resonance imaging system of claim 34 wherein a given current distribution is obtained on said transmission line elements at a given frequency by adjustment of the geometry of said transmission line elements and circuit components connected among said transmission line elements.

47. The magnetic resonance imaging system of claim 34 wherein the fields generated by the currents in said transmission line elements are superposed to create a given magnetic field configuration for use the generation of the NMR signal.

48. The magnetic resonance imaging system of claim 34 including RF power amplifiers coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal.

49. The magnetic resonance imaging system of claim 34 including at least an RF power amplifier reactively coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal, the impedance of said RF power amplifier and the impedance of said one of said transmission line elements being matched.

50. The magnetic resonance imaging system of claim 34 wherein the phases of the current in a plurality of said transmission line elements are offset so as to create an elliptically polarized magnetic field for generating and/or detecting nuclear magnetic resonance signals.

51. The magnetic resonance imaging system of claim 34 including a plurality of diodes operatively connected to a plurality of said transmission line elements for tuning the coupling between said transmission line elements.

52. The magnetic resonance imaging system of claim 34 including coupling components between one or more of said transmission line elements to allow the currents on each of said transmission line elements to be independently controlled with separate power amplifiers connected to one or more of said transmission line elements allowing a transmit field of desired spatial intensity and phase to be generated.

53. The magnetic resonance imaging system of claim 34 with individual power amplifiers connected to each transmission line element with impedance mismatches designed to allow the current of each transmission line element to be independently controlled allowing a transmit field of desired spatial intensity and phase to be generated.

54. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis; (b) a receive inductor system about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said receive inductor system; (d) a field inductor system operatively associated with said receive inductor system; (e) said receive inductor system constituting a coil having an outer surface about said axis and including a series of electrical transmission line elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface, said coil including a means for detuning said coil to prevent disturbance of the transmit fields generated by a separate transmit inductor system; (f) said coil receiving from said subject fields of magnetic resonance energy; (g) said gradient inductor system initiating perturbations in said fields and producing signals derived responsively from said perturbations; (h) said signals corresponding to spatial indicia derived from said subject.

55. The magnetic resonance imaging system of claim 54 wherein, by adjusting the distance between said transmission line elements, the interaction of the magnetic fields of said transmission line elements with an external sample can be controlled and optimized for nuclear magnetic resonance signal detection.

56. The magnetic resonance imaging system of claim 54 wherein said series of transmission line elements decrease the inductance of each transmission line element and minimize the electric fields associated therewith.

57. The magnetic resonance imaging system of claim 54 wherein said series of transmission line elements have inherent shielding.

58. The magnetic resonance imaging system of claim 50 wherein said series of transmission line elements are combined to optimize NMR signal reception.

59. The magnetic resonance imaging system of claim 50 wherein signals from said series of transmission line elements are combined to decode spatial information derived from the NMR signal.

60. The magnetic resonance imaging system of claim 50 wherein said receive inductor system consists of N transmission line elements arranged in a geometric pattern in which each of said transmission line elements is substantially equidistant from each adjacent transmission line element.

61. The magnetic resonance imaging system of claim 60 wherein said geometric pattern is circular or elliptical.

62. The magnetic resonance imaging system of claim 60 wherein said geometric pattern is flat or curved.

63. The magnetic resonance imaging system of claim 59 wherein said outer and inner conductors include additional lumped or distributed capacitive or inductive circuit components.

64. The magnetic resonance imaging system of claim 59 wherein each of said transmission line elements couples to the other of said transmission line elements through mutual inductance and capacitive coupling.

65. The magnetic resonance imaging system of claim 59 wherein distributed impedance elements are connected between certain of said transmission line elements alter the coupling therebetween.

66. The magnetic resonance imaging system of claim 59 wherein impedance elements are connected between said transmission line elements to establish interactions that establish frequency dependent relations between the currents and voltages present on certain of said transmission line elements.

67. The magnetic resonance imaging system of claim 59 wherein a=given current distribution is obtained on said transmission line elements at a given frequency by adjustment of the geometry of said transmission line elements and circuit components connected among said transmission line elements.

68. The magnetic resonance imaging system of claim 59 wherein the fields generated by the currents in said transmission line elements are superposed to create a given magnetic field configuration for use in the detection of the NMR signal.

69. The magnetic resonance imaging system of claim 59 including RF receivers coupled to at least one of said transmission line elements for transferring energy out of said coil during receive.

70. The magnetic resonance imaging system of claim 59 wherein the phases of the current in a plurality of said transmission line elements are offset so as to create an elliptically polarized magnetic field for detecting nuclear magnetic resonance signals.

71. The magnetic resonance imaging system of claim 69 including a plurality of diodes operatively connected to a plurality of said transmission line elements for tuning the coupling between said transmission line elements and said RF receivers.

72. The magnetic resonance imaging system of claim 59 including coupling elements between one or more of said transmission line elements in order to make the currents on each of said transmission line elements relatively independent allowing the signals from two or more of said transmission line elements to be optimally combined before or after image reconstruction.

73. The magnetic resonance imaging system of claim 59 with individual preamplifiers connected to each of said transmission line elements with impedance mismatches designed to allow each of said transmission line elements to operate independently

allowing the signals from two or more of said transmission line element to be optimally combined either before or after image reconstruction.

74. The magnetic resonance imaging system of claim 59 with individual preamplifier/receivers connected to each transmission line element with the independent information obtained from individual transmission line elements being used to decode spatial information regarding said subject.

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L20: Entry 1 of 22 File: PGPB Jan 30, 2003

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PUBLICATION-DATE: January 30, 2003

INVENTOR-INFORMATION:

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US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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TITLE: Multimode operation of quadrature phased array MR coil systems

PUBLICATION-DATE: January 2, 2003

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TITLE: Multimode operation of quadrature phased array MR coil systems

PUBLICATION-DATE: December 19, 2002

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INVENTOR-INFORMATION:

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PUBLICATION-DATE: September 12, 2002

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US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Hamamura, Yoshinori	Otawara-shi		JP	
Mitsui, Shinji	Nasu-gun		JP	
Yasuhara, Yasutake	Nasu-gun		JP	

US-CL-CURRENT: 324/300

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 8. Document ID: US 20010005136 A1

L20: Entry 8 of 22 File: PGPB Jun 28, 2001

PGPUB-DOCUMENT-NUMBER: 20010005136
PGPUB-FILING-TYPE: new-utility
DOCUMENT-IDENTIFIER: US 20010005136 A1

TITLE: Magnetic resonance imaging receiver/transmitter coils

PUBLICATION-DATE: June 28, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Misic, George J.	Allison Park	PA	US	

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 9. Document ID: US 6501274 B1

L20: Entry 9 of 22 File: USPT Dec 31, 2002

US-PAT-NO: 6501274
DOCUMENT-IDENTIFIER: US 6501274 B1

TITLE: Magnetic resonance imaging system using coils having paraxially distributed transmission line elements with outer and inner conductors

DATE-ISSUED: December 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ledden; Patrick	Malden	MA		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 10. Document ID: US 6396273 B2

L20: Entry 10 of 22 File: USPT May 28, 2002

US-PAT-NO: 6396273
DOCUMENT-IDENTIFIER: US 6396273 B2

TITLE: Magnetic resonance imaging receiver/transmitter coils

DATE-ISSUED: May 28, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George J.	Allison Park	PA		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 11. Document ID: US 6377044 B1

L20: Entry 11 of 22 File: USPT Apr 23, 2002

US-PAT-NO: 6377044
DOCUMENT-IDENTIFIER: US 6377044 B1

TITLE: Multi-mode receiver coils for MRI

DATE-ISSUED: April 23, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Burl; Michael	Chagrin Falls	OH		
Missal; John W.	Willoughby	OH		
Chmielewski; Thomas	Willoughby Hills	OH		

US-CL-CURRENT: 324/307; 324/309, 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 12. Document ID: US 6356081 B1

L20: Entry 12 of 22 File: USPT Mar 12, 2002

US-PAT-NO: 6356081
DOCUMENT-IDENTIFIER: US 6356081 B1

TITLE: Multimode operation of quadrature phased array MR coil systems

DATE-ISSUED: March 12, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George J.	Allison Park	PA		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 13. Document ID: US 6194900 B1

L20: Entry 13 of 22 File: USPT Feb 27, 2001

US-PAT-NO: 6194900

DOCUMENT-IDENTIFIER: US 6194900 B1

TITLE: Integrated miniaturized device for processing and NMR detection of liquid phase samples

DATE-ISSUED: February 27, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Freeman; Dominique M.	Pescadero	CA		
Swedberg; Sally A.	Palo Alto	CA		

US-CL-CURRENT: 324/321; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 14. Document ID: US 6177797 B1

L20: Entry 14 of 22 File: USPT Jan 23, 2001

US-PAT-NO: 6177797
DOCUMENT-IDENTIFIER: US 6177797 B1

TITLE: Radio-frequency coil and method for resonance/imaging analysis

DATE-ISSUED: January 23, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 15. Document ID: US 6150816 A

L20: Entry 15 of 22 File: USPT Nov 21, 2000

US-PAT-NO: 6150816
DOCUMENT-IDENTIFIER: US 6150816 A

TITLE: Radio-frequency coil array for resonance analysis

DATE-ISSUED: November 21, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 16. Document ID: US 6100694 A

L20: Entry 16 of 22 File: USPT Aug 8, 2000

US-PAT-NO: 6100694
DOCUMENT-IDENTIFIER: US 6100694 A

TITLE: Multiple-tuned bird cage coils

DATE-ISSUED: August 8, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wong; Wai Ha	San Jose	CA		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 17. Document ID: US 6040697 A

L20: Entry 17 of 22 File: USPT Mar 21, 2000

US-PAT-NO: 6040697
DOCUMENT-IDENTIFIER: US 6040697 A
**** See image for Certificate of Correction ****

TITLE: Magnetic resonance imaging receiver/transmitter coils

DATE-ISSUED: March 21, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George J.	Allison Park	PA		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 18. Document ID: US 5898306 A

L20: Entry 18 of 22 File: USPT Apr 27, 1999

US-PAT-NO: 5898306
DOCUMENT-IDENTIFIER: US 5898306 A
**** See image for Certificate of Correction ****

TITLE: Single circuit ladder resonator quadrature surface RF coil

DATE-ISSUED: April 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Haiying	Minneapolis	MN		
Truwit; Charles L.	Wayzata	MN		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 19. Document ID: US 5664568 A

L20: Entry 19 of 22 File: USPT Sep 9, 1997

US-PAT-NO: 5664568
DOCUMENT-IDENTIFIER: US 5664568 A

TITLE: Split-top, neck and head vascular array for magnetic resonance imaging

DATE-ISSUED: September 9, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Hts.	OH		
Henderson; Robert G.	Wickliffe	OH		
Elek; Robert A.	Chardon	OH		

US-CL-CURRENT: 600/422; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 20. Document ID: US 5374890 A

L20: Entry 20 of 22 File: USPT Dec 20, 1994

US-PAT-NO: 5374890
DOCUMENT-IDENTIFIER: US 5374890 A

TITLE: Simultaneous magnetic resonance imaging of multiple human organs

DATE-ISSUED: December 20, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Zou; Xueming	Chesterland	OH		
Patrick; John L.	Chagrin Falls	OH		
McNally; James M.	Chagrin Falls	OH		

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 21. Document ID: US 5370118 A

L20: Entry 21 of 22 File: USPT Dec 6, 1994

US-PAT-NO: 5370118
DOCUMENT-IDENTIFIER: US 5370118 A

TITLE: Opposed loop-pair quadrature NMR coil

DATE-ISSUED: December 6, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Vij; Kamal	New Berlin	WI		
Boskamp; Eddy B.	Menomonee Falls	WI		

US-CL-CURRENT: 600/422; 324/311, 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 22. Document ID: US 4689563 A

L20: Entry 22 of 22 File: USPT Aug 25, 1987

US-PAT-NO: 4689563
DOCUMENT-IDENTIFIER: US 4689563 A

TITLE: High-field nuclear magnetic resonance imaging/spectroscopy system

DATE-ISSUED: August 25, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Bottomley; Paul A.	Clifton Park	NY		
Edelstein; William A.	Schenectady	NY		
Hart, Jr.; Howard R.	Schenectady	NY		
Schenck; John F.	Schenectady	NY		
Redington; Rowland W.	Schenectady	NY		
Leue; William M.	Albany	NY		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Term	Documents
TWO.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	5473464
TWOES	0
TWOS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2558
TWOE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	13
"2".DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	15514481
2S.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	16495
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Search Results - Record(s) 1 through 10 of 10 returned.

☐ 1. Document ID: US 20030020475 A1

L22: Entry 1 of 10   File: PGPB   Jan 30, 2003

PGPUB-DOCUMENT-NUMBER: 20030020475
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030020475 A1

TITLE: RF coil system for an MR apparatus

PUBLICATION-DATE: January 30, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Leussler, Christoph Guenther	Hamburg		DE	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 2. Document ID: US 20020125888 A1

L22: Entry 2 of 10   File: PGPB   Sep 12, 2002

PGPUB-DOCUMENT-NUMBER: 20020125888
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020125888 A1

TITLE: Magnetic resonance imaging apparatus

PUBLICATION-DATE: September 12, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Visser, Frederik	Eindhoven		NL	
Haans, Paulus Cornelius Hendrikus Adrianus	Eindhoven		NL	
Van Den Brink, Johan Samuel	Eindhoven		NL	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 3. Document ID: US 20020040185 A1

L22: Entry 3 of 10   File: PGPB   Apr 4, 2002

PGPUB-DOCUMENT-NUMBER: 20020040185
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020040185 A1

TITLE: Systems and methods for evaluating the urethra and the periurethral tissues

PUBLICATION-DATE: April 4, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Atalar, Ergin	Columbia	MD	US	
Quick, Harald	Essen-Werden	MD	DE	
Karmarkar, Parag	Elliott City		US	

US-CL-CURRENT: 600/423

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 4. Document ID: US 20010022515 A1

L22: Entry 4 of 10 File: PGPB Sep 20, 2001

PGPUB-DOCUMENT-NUMBER: 20010022515
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20010022515 A1

TITLE: Magnetic resonance imaging apparatus

PUBLICATION-DATE: September 20, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Yamashita, Masatoshi	Otawara-shi		JP	
Ishii, Manabu	Otawara-shi		JP	
Sakakura, Yoshitomo	Nasu-gun		JP	
Takamori, Hiromitsu	Otawara-shi		JP	
Nakabayashi, Kazuto	Nasu-gun		JP	
Hamamura, Yoshinori	Otawara-shi		JP	
Mitsui, Shinji	Nasu-gun		JP	
Yasuhara, Yasutake	Nasu-gun		JP	

US-CL-CURRENT: 324/300

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 5. Document ID: US 6377044 B1

L22: Entry 5 of 10 File: USPT Apr 23, 2002

US-PAT-NO: 6377044
DOCUMENT-IDENTIFIER: US 6377044 B1

TITLE: Multi-mode receiver coils for MRI

DATE-ISSUED: April 23, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Burl; Michael	Chagrin Falls	OH		
Missal; John W.	Willoughby	OH		
Chmielewski; Thomas	Willoughby Hills	OH		

US-CL-CURRENT: 324/307; 324/309, 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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6. Document ID: US 6150816 A

L22: Entry 6 of 10 File: USPT Nov 21, 2000

US-PAT-NO: 6150816
DOCUMENT-IDENTIFIER: US 6150816 A

TITLE: Radio-frequency coil array for resonance analysis

DATE-ISSUED: November 21, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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7. Document ID: US 5898306 A

L22: Entry 7 of 10 File: USPT Apr 27, 1999

US-PAT-NO: 5898306
DOCUMENT-IDENTIFIER: US 5898306 A
**** See image for Certificate of Correction ****

TITLE: Single circuit ladder resonator quadrature surface RF coil

DATE-ISSUED: April 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Haiying	Minneapolis	MN		
Truwit; Charles L.	Wayzata	MN		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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8. Document ID: US 5664568 A

L22: Entry 8 of 10 File: USPT Sep 9, 1997

US-PAT-NO: 5664568
DOCUMENT-IDENTIFIER: US 5664568 A

TITLE: Split-top, neck and head vascular array for magnetic resonance imaging

DATE-ISSUED: September 9, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Hts.	OH		
Henderson; Robert G.	Wickliffe	OH		
Elek; Robert A.	Chardon	OH		

US-CL-CURRENT: 600/422; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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9. Document ID: US 5370118 A

L22: Entry 9 of 10 File: USPT Dec 6, 1994

US-PAT-NO: 5370118
DOCUMENT-IDENTIFIER: US 5370118 A

TITLE: Opposed loop-pair quadrature NMR coil

DATE-ISSUED: December 6, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Vij; Kamal	New Berlin	WI		
Boskamp; Eddy B.	Menomonee Falls	WI		

US-CL-CURRENT: 600/422; 324/311, 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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10. Document ID: US 4689563 A

L22: Entry 10 of 10 File: USPT Aug 25, 1987

US-PAT-NO: 4689563
DOCUMENT-IDENTIFIER: US 4689563 A

TITLE: High-field nuclear magnetic resonance imaging/spectroscopy system

DATE-ISSUED: August 25, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Bottomley; Paul A.	Clifton Park	NY		
Edelstein; William A.	Schenectady	NY		
Hart, Jr.; Howard R.	Schenectady	NY		
Schenck; John F.	Schenectady	NY		
Redington; Rowland W.	Schenectady	NY		
Leue; William M.	Albany	NY		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Term	Documents
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ENCODCE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
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L22: Entry 9 of 10 File: USPT Dec 6, 1994

DOCUMENT-IDENTIFIER: US 5370118 A
TITLE: Opposed loop-pair quadrature NMR coil

Abstract Text (1):
A quadrature local coil includes two coil sets placed on opposite sides of the patient, each coil set having a single loop and a split loop so as to be sensitive to quadrature components of a flux field substantially centered between the coil sets. Signals are developed from the loops in a manner to reduce current flow in the loops preventing coupling of the opposing loops and the degradation of the signal. The signals may be summed to produce a single signal of improved signal-to-noise ratio.

Number of Drawing Sheets (1):
4

Brief Summary Text (3):
The field of the invention is magnetic resonance imaging (MRI) and, in particular, local coils for use in receiving MRI signals.

Brief Summary Text (5):
A. MRI Imaging

Brief Summary Text (6):
In MRI, a uniform magnetic field B.sub.0 is applied to an imaged object along the z-axis of a Cartesian coordinate system, the origin of which is approximately centered within the imaged object. The effect of the magnetic field B.sub.0 is to align the object's nuclear spins along the z-axis.

Brief Summary Text (9):
Hydrogen, and in particular the nucleus (protons), because of its relative abundance in biological tissue and the properties of its nuclei, is of principle concern in such imaging. The value of the gyromagnetic ratio .gamma. for protons is 4.26 kHz/gauss and therefore, in a 1.5 Tesla polarizing magnetic field B.sub.0, the resonant or Larmor frequency of protons is approximately 63.9 MHz.

Brief Summary Text (10):
In a typical imaging sequence for an axial slice, the RF excitation signal is centered at the Larmor frequency .omega. and applied to the imaged object at the same time as a magnetic field gradient G.sub.z is applied. The gradient field G.sub.z causes only the nuclei, in a slice with a limited width through the object along an x-y plane, to have the resonant frequency .omega. and to be excited into resonance.

Brief Summary Text (11):
After the excitation of the nuclei in this slice, magnetic field gradients are applied along the x and y axes. The gradient along the x-axis, G.sub.x, causes the nuclei to precess at different frequencies depending on their position along the x-axis, that is, G.sub.x spatially encodes the precessing nuclei by frequency. The y axis gradient, G.sub.y, is incremented through a series of values and encodes the y position into the rate of change of phase of the precessing nuclei as a function of gradient amplitude, a process typically referred to as phase encoding.

Brief Summary Text (12):
A weak nuclear magnetic resonance generated by the precessing nuclei may be sensed by the RF coil and recorded as an NMR signal. From this NMR signal, a slice image

may be derived according to well known reconstruction techniques. An overview of NMR image reconstruction is contained in the book "Magnetic Resonance Imaging, Principles and Applications" by D. N. Kean and M. A. Smith.

Brief Summary Text (14):
The quality of the image produced by MRI techniques is dependent, in part, on the strength of the NMR signal received from the precessing nuclei. For this reason, it is best to use an independent RF receiving coil placed in close proximity to the region of interest of the imaged object in order to improve the strength of this received signal. Such coils are termed "local coils" or "surface coils". The smaller area of the local coil permits it to accurately focus on NMR signals from the region of interest. Further, the RF energy of the field of such a local coil is concentrated in a smaller volume giving rise to improved signal-to-noise ratio in the acquired NMR signal.

Brief Summary Text (15):
The signal-to-noise ratio of the NMR signal may be further increased by orienting two coil pairs at 90.degree. angles about the imaged object so that each detects RF energy along one of a pair of mutually perpendicular axes. This technique is generally known as quadrature detection and the signals collected are termed quadrature signals.

Brief Summary Text (16):
The outputs of the quadrature coil pairs are combined so as to increase the strength of the received signal according to the simple sum of the output signals from the coils. The strength of the uncorrelated noise component of these signals, however, will increase only according to the square root of the sum of the squares of the noise components. As a result, the net signal-to-noise ratio of the combined quadrature signals increases by approximately .sqroot.2 over the signal-to-noise ratio of the individual signal.

Brief Summary Text (17):
The quadrature orientation of the two coils introduces a 90.degree. phase difference between the NMR signals detected by these coils. Therefore, combining the outputs from the two quadrature coils, to achieve the above described signal-to-noise ratio improvement, requires that one signal be shifted to have the same phase as the other signal so that the amplitudes of the signals simply add.

Brief Summary Text (18):
Such phase shifting and combining is typically accomplished by means of a hybrid network. Hybrid networks are four-port networks known in the art and having the property that when the four ports are properly terminated, energy input to two of the ports, with the proper relative phase angles, will be combined at one of the remaining two ports. The antenna coils are attached to two of the ports and the output lead is attached to a third port and produces the sum of the signals from the antenna coils, one being shifted so that they add in-phase. The remaining uncommitted port is connected to a termination resistor.

Brief Summary Text (19):
As used herein, the term quadrature coil and quadrature signal, will refer to the detecting of the NMR signal along multiple axes and combining the signals so collected, with the appropriate phase shifts to produce a signal of improved signal-to-noise ratio.

Brief Summary Text (21):
One method of constructing a local coil is the "bird cage" construction in which two conductive loops are spaced apart along a common longitudinal axis and interconnected by a series of regularly spaced longitudinal connectors. The impedance of the loops and of the longitudinal conductors is adjusted so that the coil may be excited into resonance by a rotating transverse magnetic field at the Larmor frequency. A quadrature signal may be obtained by monitoring the current through two longitudinal conductors spaced at 90.degree. around the periphery of the loops. Such coils are described in detail in U.S. Pat. Nos. 4,680,548, 4,692,705, 4,694,255 and 4,799,016.

Brief Summary Text (22):
For certain imaging tasks, particularly imaging of the neck, the use of a volumetric coil such as a bird cage is impractical because of the diameter of the coil required to clear the patient's shoulders so that the region of sensitivity of the coil may

be centered about the neck. The use of "cut outs" in one of the conductive loops of the bird cage to fit about the shoulders of the patient has proven unsatisfactory.

Brief Summary Text (24):
The use of volumetric local coils of conventional bird cage or other quadrature design may be undesirably constraining to the patient who must be surrounded by the relatively small volume of the tubular coil within the magnet bore. Often, in order that the local coil may be conveniently located on the patient, it is desirable that the quadrature local coil be opened and then closed about the desired anatomy. Coils that may be thus opened are not easily implemented with the bird cage design.

Brief Summary Text (25):
It is known, therefore, for certain imaging applications, such as the imaging of the spine, to construct a quadrature local coil on a one-sided cradle to be attached to the upper surface of the patient support table so that the patient may simply lie on top of the coil and so that the coil structure is not unduly constraining. Such open coils are termed "planar" coils to distinguish them from "whole volume" coils such as might be constructed of opposed saddle coils or solenoids or a birdcage. The prior art has recognized the desirability of a quadrature, planar coil. See, for example, U.S. Pat. No. 5,030,915, issued Jul. 9, 1991 to Boskamp, hereby incorporated by reference.

Brief Summary Text (28):
A quadrature planar coil may be readily placed, without interference from the patient's shoulders, in the region of the neck. Unfortunately, its region of sensitivity is closely concentrated near the plane of the coils which may not be suitable for imaging structures in the neck region removed from that plane.

Brief Summary Text (29):
For this reason, a specialized two-part saddle-shape neck coil such, as is taught in U.S. Pat. No. 5,221,902 assigned to the same assignee as the present invention and hereby incorporated by reference, was designed which permits the accommodation of the patient's shoulders allowing centering the useful volume of the coil in the neck region.

Brief Summary Text (32):
The present invention provides a quadrature local coil suitable for imaging portions of the body such as the neck where a bird cage type coil would be impractical and a planar quadrature coil would present too shallow of an imaging depth.

Brief Summary Text (33):
Generally two quadrature "planar" coils are opposed about the volume to be imaged. Specialized interface circuitry is used to eliminate the coupling, or loss of isolation, that would otherwise be expected of this combination, and to combine the signals from these two pairs.

Brief Summary Text (34):
Specifically, the local coil comprises a first coil set positioned adjacent to the imaging volume, the set having a first coil with a first reception pattern which couples with an RF magnetic field having a first orientation within the imaging volume. The first coil produces a first signal. The first coil set also has a second coil with at least one diametric conductor to divide the second coil into a pair of loops having a second reception pattern which couples to an RF magnetic field of a second orientation within the imaging volume to produce a second signal. The second orientation is substantially 90.degree. from the first orientation to produce a quadrature sensitivity. This first coil pair is substantially opposed about the imaging volume by a second coil pair. The second coil pair has a third coil with a reception pattern of the first orientation to produce a third signal and fourth coil having at least one diametric conductor to divide the fourth coil into a pair of loops having the second reception pattern to produce a fourth signal.

Brief Summary Text (35):
The coil may include combiners for combining the first and third signals and for combining the second and fourth signals into combined output signals. Another combiner combines the output signals with a relative phase shift in one signal to produce a quadrature output.

Brief Summary Text (36):
It is one object of the invention to provide a quadrature coil having good

accessibility and a well located imaging area by combining opposed planar style quadrature coils. Each half of the coil is electrically independent and may be hinged or otherwise opened with respect to the other half to provide good patient access. When the two halves are closed in opposition, the imaging area may be approximately centered between the two halves rather than proximate to one, as would be the case with a single planar coil.

Drawing Description Text (2):
FIG. 1 is a perspective view of a housing supporting the local coil of the present invention as adapted for imaging of the neck and showing the posterior and anterior coils in the open position;

Drawing Description Text (3):
FIG. 2 is an elevation in cross-section along a midsagittal plane showing positioning of the local coil of FIG. 1 on a patient;

Drawing Description Text (5):
FIGS. 4(a) and (b) are cross-sections of the conductors of the present invention showing magnetic coupling between the opposing corresponding coils of an anterior and posterior coil set.

Drawing Description Text (6):
FIG. 5 is a schematic diagram of the coils of the coil sets of FIG. 1 showing combining of the signals from each coil; and

Drawing Description Text (7):
FIG. 6 is a figure similar to that of FIGS. 4(a) and (b) showing quadrature detection of the two coil pairs in a region of interest centered between the coil pairs.

Detailed Description Text (2):
Referring to FIGS. 1 and 2, quadrature coil 10 of the present invention, when configured for use in neck imaging, includes opposed anterior and posterior coil sets 12 and 14. The posterior coil set 14 is supported by a generally horizontal, planar base 16 whereas the anterior coil set 12 is held away from the base 16 by an extension tower 18 projecting perpendicularly upward from the horizontal surface of the base 16. The base 16 has on its lower surface a number of downwardly extending arc shaped ribs 73 that fit against the concave upper surface of the MRI table 75 (shown in FIG. 1) to stiffen the base 16 and to provide additional support for the base 16 against the table 75.

Detailed Description Text (4):
Referring to FIG. 2, when the local coil 10 is in use, the patient's head rests back against the upper surface of the base 16 with the patient's frontal plane generally parallel to the surface of the base 16. Left and right medially extending wedges 20 rise from the upper surface of the base 16. The wedges are symmetrically opposed about the medial axis 22 and support and position a trough shaped cushion 24 that cradles either side of the patient's neck and head when the patient is positioned in the coil 10. Held within the trough is a transverse arched foam pad 26 which supports the back of the patient's neck and tips the patient's head upward on the base 16. The foam pad 26 also covers an arcade conductor to be described below.

Detailed Description Text (5):
Flat foam cushions 28 are positioned against the base 16 above and below the arched foam pad 26 along the medial axis 22, to support the back of the patient's head and shoulders.

Detailed Description Text (7):
As mentioned, the left sidebar 36 of the anterior coil set 12 is attached to a hinge (not shown) which permits the chest arch 32 to be retracted away from the point of patient entry when the anterior coil set 12 is moved to the open position, thus improving the access for the patient who normally lays back against the base 16 and whose head enters the coil set at a relatively steep angle. The above structural components are fabricated from a non-magnetic, non-conductive, polymeric material to reduce their interaction with the magnetic and electrical fields of the MRI equipment. The mechanism of opening and adjusting the anterior coil set 12 is described in detail in U.S. Pat. No. 5,166,618 entitled: "NMR Neck Coil with Passive Decoupling" hereby incorporated by reference.

Detailed Description Text (8):
Referring to FIGS. 1, 2 and 3(a) through 3(d), the anterior and posterior antenna coil sets 12 and 14 each include two distinct antenna loops. The anterior coil set 12 has a single loop 40 and a split loop 42. Electrically the split loop 42 differs from the single loop in that it is effectively bifurcated by a conductor 41 along the medial axis so as to produce two electrical loops. This additional bifurcating conductor 41 of the split loop 42 passes along the mid-sagittal anterior beam 33. Mechanically, the split loop 42 is formed from a single conductor twisted to a figure-eight shape with the two loops of the figure-eight abutting along the medial axis 22 to form the bifurcating conductor 41.

Detailed Description Text (10):
Referring still to FIGS. 3(a) through (d), the posterior coil set 14 also comprise a single loop 44 and a split loop 46, the latter differing from the former again by the introduction of a bifurcating conductor 45 which follows the arch of the neck of the patient beneath the arched foam pad 26. Like the split loop 42 of the anterior coil set 12, the split loop 46 is essentially a single conductor formed in a FIG. 8 with the conductors of each loop of the 8 forming the bifurcating conductor 45 in the midsagittal posterior arch 35. With the exception of the conductors through this midsagittal posterior arch 35, the conductors of the single loop 44 and the split loop 46 follow generally the support structure previously described having arcade portions which rise on either side of the patient's neck, when the patient is positioned on the coil set 14, and that fit into the channels in the left and right wedges 20. These arcade portions are connected by conductor segments substantially parallel to the plane of the base 16 and contained with the base 16 in a protective housing 68.

Detailed Description Text (11):
Referring to FIG. 5, each of the loops 40, 42, 44 and 46 is cut at two points spaced equally about the loop. These cuts are bridged by capacitors 50 which together with the intrinsic inductance of the conductors of the loop, serve to tune the loop into resonance at a frequency equal to the Larmor frequency of the precessing nuclei whose NMR signals are to be detected.

Detailed Description Text (13):
For single loop 40, the capacitor 50' is bridged by a back-to-back diode network 52 which serves to limit the voltage of the signal developed across capacitor 50' to approximately 7/10ths of a volt peak-to-peak. Voltages of greater than this threshold would otherwise be obtained during the excitation of the nuclei by the RF excitation coil. These high voltages may also be avoided by active decoupling of the loops, as will be described, but back-to-back diodes 52 provide a second means of decoupling the single loop 40 from the excitation signal in the event that active decoupling is not provided.

Detailed Description Text (14):
One side of capacitor 50' also connects to inductor 56 which in turn is connected to a phase shifter 58. This phase shifter may be a cable or a discrete network. The voltage across capacitor 50' is carried via the phase shifter 58 to preamplifier 60 which at its output produces a signal 62 proportional to the signal received by the single loop 40. Preamplifier 60 is low noise preamplifier such as are well known in the art.

Detailed Description Text (15):
Phase shifter 58 provides an impedance transformation of the input impedance of the preamplifier 60 so as to present an impedance joining the inductor 56 across the capacitor 50'. This joining impedance provides an effective shunting of capacitor 50' by an inductance whose value is determined by the series combination of the inductor 56 and the transformed input impedance of preamplifier 60.

Detailed Description Text (17):
As mentioned, loop 40 may also be actively decoupled by means of a well known decoupling circuit 64 (shown only as a block) which includes a diode that may be forward biased by an independent source of DC power (not shown) to connect an inductor in the decoupling block 64 across capacitor 50. The inductor is sized to detune the loop 40 when the diode is forward biased and during the RF excitation of the nuclear spins, thus preventing excessive signals from reaching and damaging the preamplifier 60. The source of DC power must be timed to the RF excitation and is generally provided by the manufacturer of the MRI equipment.

Detailed Description Text (19):
Referring now to FIG. 6, the single loops 40 and 44 will have an induced voltage as a result of flux 72 along a y-axis line passing through both of their centers. The polarity of the connections of single loops 40 and 44 to their respective preamplifiers 60 is adjusted such that flux line 72 passing through both loops 40 and 44 in the same direction produce signals 62 and 70 that are in the same phase. Within a region of interest 74, flux lines 72 are parallel to the y-axis and thus provide a y-axis sensitivity for quadrature detection.

Detailed Description Text (21):
As described above, unlike prior art planar quadrature coils, each of the loops of the coil sets of the present invention are not intrinsically isolated from the other loops. Referring to FIG. 4(a), absent the decoupling effect of the connection of preamplifiers 60, described above, a cyclic current passing through the single loop 40 generates flux lines 72' which will be intercepted by single loop 44 to induce current in that single loop 44. Thus, the loops 40 and 44 are not isolated, and this has the effect of detuning the loops and thus decreasing the signal-to-noise ratio of their combined signal.

Detailed Description Text (22):
In contrast, these same flux lines 72' passing between the two loops of split loop 42 in intervals 78 and 80, and split loop 46 induce countervailing currents which cancel within the split loops 46 and 42 and thus in general, each split loop 42 or 46 is intrinsically isolated from either single loop 40 and 44. Referring further to FIG. 4(b), a current flowing in either of split loops 42 or 46 produces countervailing flux lines 76 which pass in both directions in equal density through loops 40 and 44 (in region 77, for example). Thus, the single loops 40 and 44 are also intrinsically isolated from the split loops 42 and 46.

Detailed Description Text (24):
Referring again to FIG. 5, the signals 66 and 68, from the split loops 42 and 46 respectively, are summed by means of a combining network 82 providing the necessary combining and impedance matching of the signal 66 and 68 according to the polarities described with respect to FIG. 6. The combining network 82 may be a hybrid combiner such as a well known in the art. The output of the combiner 82 yields the x-axis of the quadrature signal 84.

Detailed Description Text (25):
Likewise, signal 62 and 70 from single loop 40 and 44 are combined by a second combiner 86 identical to that of 82. The output of the combiner 86 produces the y-axis component of the quadrature signal 88. Signals 88 and 84 are in turn provided to a 90.degree. hybrid combiner 90 which combine them to produce the NMR signal 92 having improved signal-to-noise ratio.

Detailed Description Text (26):
The above description has been that of a preferred embodiment of the present invention. It will occur to these who practice the art that many modifications may be made without departing from the spirit and scope of the invention. For example, the two loops of the posterior cell 14 may be physically separate, provided they are substantially adjacent and have their signals combined as described. Clearly, the position of the anterior and posterior loops 44 and 46 may be reversed. Further, it will be apparent from this description that the present coil design may be used not only in receive only coils but in coils that also transmit the exciting RF MRI pulse. In order to apprise the public of the various embodiments that may fall within the scope of the invention, the following claims are made.

CLAIMS:

1. An NMR probe for obtaining an NMR signal from precessing nuclei within an imaging volume, the probe comprising:

a first coil set positioned adjacent to the imaging volume including:

(a) a first coil having a first reception pattern which couples to an RF magnetic field of a first orientation within the imaging volume to produce a first signal;

(b) a second coil having at least one diametric conductor to divide the second coil into a pair of loops having a second reception pattern which couples to an RF magnetic field of a second orientation within the imaging volume to produce a second

signal, the second orientation having an angular separation from the first orientation of substantially 90.degree. measured in the direction of the precession of nuclei; and

a second coil set opposed substantially symmetrically to the first coil set about the imaging volume including:

(c) a third coil having a third reception pattern which couples to the RF magnetic field of the first orientation within the imaging volume to produce a third signal;

(d) a fourth coil having at least one diametric conductor to divide the fourth coil into a pair of loops having the second reception pattern which couples to the RF magnetic field of the second orientation within the imaging volume to produce a fourth signal.

2. The NMR probe of claim 1 including additionally;

a first combiner means receiving the first and third signal for combining the first and third signals into a combined output signal;

a second combiner means receiving the second and fourth signals for combining the second and fourth signals into a combined output signal; and

a third combiner means for combining the output signals from the first and second combiner means so that the output signal from the first combiner means is shifted in phase by 90.degree. with respect to the output signal from the second combiner means.

3. The NMR probe of claim 1 wherein the first coil is saddle shaped for embracing the frontal half of a patient's neck and wherein the second coil is saddle shaped for embracing the posterior portion of the patient's neck.

4. The NMR probe of claim 3 wherein the diametric conductors of the second and fourth coils are positioned to lie in a midsagittal plane when the first and second coils sets are positioned on the patient.

5. The NMR probe of claim 1 including an electrical network blocking current flow in the coils at a signal frequency.

6. The NMR probe of claim 1 including a capacitive element, an inductive element, a phase shifting network and a preamplifier and wherein at least one of the first, second, third, and fourth signals is taken across the capacitive element in series with the respective coil and received by the inductive element connected via the phase shifting network to the preamplifier together to shunt the capacitive element with an impedance; and

wherein the capacitive element and the shunting impedance form a parallel resonant circuit at a frequency of the NMR signal.

DOCUMENT-IDENTIFIER: US 6377044 B1
TITLE: Multi-mode receiver coils for MRI

Abstract Text (1):
A magnetic resonance apparatus includes a multi-mode receiver assembly which facilitates operation in both a quadrature combination mode and phased array mode. The multi-mode receiver assembly includes a receiver coil assembly (30) comprising a first RF coil assembly (32) and a second RF coil assembly (34). A signal combining circuit, which includes a switch means, performs at least one of combining and splitting magnetic resonance signals received by the first and second RF coil assemblies (30, 32). The application of a DC bias potential to the switch means switches the multi-mode receiver assembly into the quadrature combination mode in which the received magnetic resonance signals are phase shifted and combined into a quadrature signal and an anti-quadrature signal. The absence of a DC bias potential to the switch means switches the multi-mode receiver assembly into the phased array mode in which the received magnetic resonance signals are phase shifted and passed individually to corresponding receivers. The multi-mode capability of the receiver assembly allows an operator to switch from a quadrature mode, which is provides faster reconstruction, to a phased array mode, which provides better image quality, within a single examination.

Number of Figures (1):
4

Brief Summary Text (2):
The present invention relates to the magnetic resonance arts. It finds particular application in conjunction with medical imaging and will be described with particular reference thereto. It is to be appreciated, however, that the invention may find further application in quality control inspections, spectroscopy, and the like.

Brief Summary Text (3):
Conventionally, magnetic resonance systems generate a strong, temporally constant main magnetic field, commonly denoted B.sub.0, in a free space or bore of a magnet. This main magnetic field polarizes the nuclear spin system of an object. Nuclear spins of the object then possess a macroscopic magnetic moment vector preferentially aligned with the direction of the main magnetic field. In a superconducting annular magnet, the B.sub.0 magnetic field is generated along the longitudinal axis of the cylindrical bore, which is typically assigned to be the z-axis. In an open system, the B.sub.0 magnetic field is typically oriented vertically between a pair of pole pieces, which is again assigned to be the z-axis.

Brief Summary Text (4):
To generate a magnetic resonance signal, the polarized spin system is excited at resonance by applying a radio frequency (RF) magnetic field B.sub.1, with a vector component perpendicular to that of the B.sub.0 field. In a transmission mode, the radio frequency coil is pulsed to tip the magnetization of the polarized sample away from the z-axis. As the magnetization precesses around the z-axis, the precessing magnetic moment generates a magnetic resonance signal at the Lamor frequency which is received by the same or another radio frequency coil in a reception mode.

Brief Summary Text (5):
Traditionally, RF receiver coils have been utilized with magnetic resonance imaging and spectroscopy equipment in either quadrature mode or phased array mode. Quadrature coils typically include at least two coils or coil arrays which view the

same region of interest, but are sensitive to signals 90.degree. out of phase, such as a vertical field and a horizontal field. Analogously , birdcage coils, which are circularly polarized, have taps for two 90.degree. out of phase output signals. Typically, the 90.degree. offset signals from the two coils or coil arrays are connected to an analog phase shifting circuit which causes both signals to have the same phase. Phase shifting and summing the signals typically provides a signal to noise improvement of about the square root of 2. Quadrature mode is preferable where a limited number of channels exists and speed of reconstruction is important.

Brief Summary Text (6):
Alternately, the receiver coils may be operated in a phased array mode in which the 90.degree. offset signals are each forwarded individually to separate receivers. Operation in phased array mode is preferable where improved image quality is important, such as in transverse or coronal scans. Prior art coils either make the quadrature combination on the coil in quadrature mode or output multiple signals to multiple receivers in phased array mode without the ability to switch from one mode to the other.

Brief Summary Text (9):
In accordance with one aspect of the present invention, a magnetic resonance apparatus includes a main magnet which generates a main magnetic field through an examination region. A radio frequency (RF) transmitter coil positioned about the examination region excites magnetic resonance dipoles therein. An RF transmitter drives the RF transmitter coil. A multi-mode RF receiver coil assembly receives magnetic resonance signals from the resonating dipoles and at least two receivers receive and demodulate output signals from the receiver coil assembly. The receiver coil assembly includes at least one first RF coil which is sensitive to a magnetic field along a first axis. The receiver coil assembly further includes at least one second RF coil which is sensitive to magnetic fields along a second axis which is orthogonal to the first axis. A signal combining circuit which is operatively connected to the first and second RF coils has a quadrature combining mode in which it quadrature combines signals received by the first and second RF coils and a phased array mode in which it passes signals received by the first and second RF coils to corresponding receivers without combining the signals. A switch assembly is connected to the signal combining circuit. The switch assembly switches the combining circuit between the quadrature combining mode and the phased array mode.

Brief Summary Text (10):
A multi-mode magnetic resonance method includes generating a main magnetic field through an examination region and transmitting RF signals into the examination region to induce magnetic resonance in nuclei. The induced magnetic resonance signals are received using a first RF coil and a second RF coil. The received magnetic resonance signals are phased shifted. One of a quadrature combination mode and a phased array mode is selected. In the quadrature combination mode, the phased shifted received magnetic resonance signals are combined, while in the phased array mode, the received magnetic resonance signals are passed uncombined. The received magnetic resonance signals are demodulated and reconstructed into an image representation.

Brief Summary Text (11):
In accordance with another aspect of the present invention, a multi-mode RF assembly for use in a magnetic resonance apparatus includes a first RF coil assembly comprising at least one RF coil which is sensitive to a magnetic field along a first axis to generate a first resonance signal. A second RF coil assembly comprising at least one RF coil is sensitive to a magnetic field along a second axis which is orthogonal to the first axis to generate a second resonance signal 90.degree. out of phase from the first resonance signal. A phase shift circuit shifts a relative phase of the first and second resonance signals by 90.degree.. A signal combining circuit combines the phase shifted first and second resonance signals. A switch assembly switches between outputting a combined signal and the first and second resonance signals.

Brief Summary Text (12):
In accordance with another aspect of the present invention, a method of quadrature operation in a magnetic resonance apparatus includes generating a temporally constant magnetic field through an examination region and transmitting RF signals into the examination region to induce magnetic resonance in nuclei. Induced magnetic resonance signals are detected in quadrature using a quadrature coil assembly. The detected quadrature signals are phase-shifted by 90.degree. and combined into a

quadrature signal and an anti-quadrature signal using a quadrature adder. The quadrature and anti-quadrature signals are transferred to a pair of receivers and reconstructed into an image representation.

Brief Summary Text (13):
One advantage of the present invention is that it provides switching between quadrature combination mode and phased array mode depending on the type of examination.

Brief Summary Text (14):
Another advantage of the present invention is that it uses the anti-quadrature signal from a quadrature combiner to improve image quality.

Brief Summary Text (16):
Yet another advantage of the present invention resides in use of phased array mode for applications which require better image quality.

Drawing Description Text (3):
FIG. 1 is a diagrammatic illustration of a magnetic resonance system in accordance with the present invention;

Drawing Description Text (4):
FIG. 2 is a schematic illustration of a switchable combination circuit in accordance with the present invention;

Drawing Description Text (5):
FIG. 3 is a schematic illustration of a multimode combination circuit in accordance with the present invention;

Drawing Description Text (6):
FIG. 4 is a preferred multi-mode receiver coil assembly in accordance with the present invention.

Detailed Description Text (2):
With reference to FIG. 1, a main magnetic field control 10 controls superconducting or resistive magnets 12 such that a substantially uniform, temporally constant main magnetic field B.sub.0 is created along a z-axis through an examination region 14. Although a bore-type magnet is illustrated in FIG. 1, it is to be appreciated that the present invention is applicable to open or vertical field magnetic systems as well. A magnetic resonance sequence applies a series of radio frequency (RF) pulses B.sub.1 and magnetic field gradient pulses to invert or excite magnetic spins, induce magnetic resonance, refocus magnetic resonance, manipulate magnetic resonance, spatially and otherwise encode the magnetic resonance, to saturate spins, and the like to generate magnetic resonance imaging and spectroscopy sequences. More specifically, gradient pulse amplifiers 20 apply current pulses to selected ones or pairs of whole body gradient coils 22 to create magnetic field gradients along x, y, and z-axes of the examination region 14. Radio frequency transmitters 24, 26, preferably digital, transmit radio frequency pulses or pulse packets to a whole-body RF birdcage coil 28 to generate the B.sub.1 radio frequency fields within the examination region. A typical radio frequency pulse is composed of a packet of immediately contiguous pulse segments of short duration which, taken together with each other and any applied gradients, achieve the selected magnetic resonance manipulation. The RF pulses are used to saturate, excite resonance, invert magnetization, refocus resonance, or manipulate resonance in selected portions of the examination region. For whole-body applications, the resonance signals are commonly picked up in quadrature by the whole-body RF birdcage coil 28.

Detailed Description Text (3):
Local coils are commonly placed contiguous to selected regions of the subject for receiving induced magnetic resonance signals from the selected regions. In the embodiment of FIG. 1, a local radio frequency coil 30 includes a planar loop coil 32 and a Helmholtz pair 34. In this configuration, the radio frequency loop coil 32 is primarily sensitive magnetic field components along a first vertical axis, while the Helmholtz pair 34 is primarily sensitive to magnetic field components along a second horizontal axis, which is orthogonal to the first axis. It is to be appreciated that other specialized RF coils, such as a birdcage coil or butterfly and loop combination, may be utilized as well for receiving magnetic resonance signals having a 90.degree. phase shifted relationship. The loop coil 32 and Helmholtz pair 34 are connected with a pair of amplifiers 36, 38. The amplified received resonance signals

are conveyed to a combination circuit 40, which includes a combiner 42 and a switch assembly 44.sub.1, 44.sub.2 for operation in a quadrature mode or a phased array mode based on the desired application. The workings of the combination circuit and switch are described more fully below. Preferably, the amplifiers 36, 38 and the combination circuit are mounted on the coil 30. The resultant radio frequency signals are demodulated by corresponding receivers 46, 48.

Detailed Description Text (4):
A sequence control processor 50 controls the gradient pulse amplifiers 20 and the transmitters 24, 26 to generate any of a plurality of magnetic resonance imaging and spectroscopy sequences, such as echo-planar imaging, echo-volume imaging, gradient and spin echo imaging, fast spin echo imaging, and the like. For the selected sequence, the receivers 46, 48 receive a plurality of magnetic resonance signals in rapid succession following RF excitation pulses. Analog-to-digital converters 52, 54, which are preferably incorporated into the receivers 46, 48 convert each magnetic resonance signal to a digital format. The analog-to-digital converters 52, 54 are disposed between the radio frequency receiving coils and the receivers for digital receivers and are disposed downstream (as illustrated) from the receivers for analog receivers. Ultimately, the demodulated radio frequency signals received are reconstructed into an image representation by a reconstruction processor 60 which applies a two-dimensional Fourier transform or other appropriate reconstruction algorithm. The image may represent a planar slice through the patient, an array of parallel planar slices, a three-dimensional volume, or the like. The image is then stored in an image memory 62 where it is accessed by a display 64, such as a video monitor, active matrix monitor, or liquid crystal display, which provides a human-readable display of the resultant image.

Detailed Description Text (5):
With reference to FIG. 2 and continued reference to FIG. 1, the combination and switch circuit 40 receives amplified magnetic resonance signals from the RF coils 32, 34 which have a substantially 90.degree. phase relationship. The combination and switch circuit includes a switch assembly 44, 442 for switching between a quadrature mode or a phased array mode. With the switches in a closed or short circuit position, the combination circuit operates in quadrature mode. The two signals are phase shifted 90.degree. relative to each other and combined using a standard combiner 42. More specifically, the signals that are shifted into phase alignment and summed become the quadrature output at channel 1, while anti-quadrature signals become the output at channel 2. With the switches in an open circuit position, the combination circuit operates in a phased array mode in which the two signals are delayed, but otherwise unaltered, and passed separately to the receivers for further signal processing.

Detailed Description Text (6):
With reference to FIG. 3 and continued reference to FIG. 2, the switching function of the switch assembly 44.sub.1, 44.sub.2, which is incorporated into the combination and switch circuit 40, is preferably achieved using a pair of PIN diodes 80, 82, as shown. The switch assembly switches between the quadrature mode and the phased array mode based on the presence or absence of a DC biasing potential at channel 2. The presence of a DC biasing potential at channel 2 forward biases the PIN diodes such that the two phase shifted signals are combined. In contrast, without a DC bias at channel 2, the PIN diodes prevent current flow, acting as open circuits, such that the two phase shifted signals are passed separately to the receivers at channel 1 and channel 2, respectively. Those skilled in the art will appreciate that this switching capability allows an operator to switch modes, depending on desired application, within a single examination. For example, for applications which require greater reconstruction speed, the operator would select the quadrature combination mode. For applications which require better image quality, the operator would select phased array mode.

Detailed Description Text (7):
In the quadrature or combination mode, the combination circuit outputs to the receivers 46, 48 a quadrature combined signal on channel 1 and an anti-quadrature signal on channel 2. It is to be appreciated that the anti-quadrature signal is approximately one-half the magnitude of the quadrature signal. The quadrature and anti-quadrature signals are then passed to separate receivers where they are demodulated and reconstructed into two image representations, as described above. In another embodiment, the quadrature and anti-quadrature image representations are then added or averaged, improving the signal-to-noise ratio.

Detailed Description Text (8):
While the present invention has been described with reference to magnetic resonance systems having two RF coils and two channels, artisans will appreciate that the present invention is applicable to magnetic resonance systems having four or more RF coils and channels as well. For example, with reference to FIG. 4, a butterfly coil or Helmholtz coil 100.sub.1 for one mode is curved to follow the contour of a patient's neck, while a flat loop coil 102.sub.1 for another mode is arrayed for imaging the spine. An array of paired butterfly coils 100.sub.2, 100.sub.3, 100.sub.4 and an array of flat loop or ladder coils 102.sub.2, 102.sub.3, 102.sub.4 are arrayed and mounted in a flat structure or the patient couch for additional imaging applications, such as a spine application, in a multi-channel configuration. In a neck imaging application, at least one butterfly and one loop or ladder coil are combined in quadrature as a single channel, leaving the remaining coils and channels free for spine imaging applications. Alternately, the coils in the head piece can be conveyed to separate receiver channels.

CLAIMS:

1. A magnetic resonance apparatus having a main magnet which generates a main magnetic field through an examination region, a radio frequency (RF) transmitter coil positioned about the examination region such that it excites magnetic resonance dipoles therein, an RF transmitter which drives the RF transmitter coil, a multi-mode RF receiver coil assembly which receives magnetic resonance signals from the resonating dipoles, said multi-mode RF receiver coil assembly being switchable between a quadrature combining mode and a phased array mode without a change in RF coil structure, and at least two receivers which receive and demodulate output signals from the multi-mode receiver coil assembly, the multi-mode receiver coil assembly comprising:

at least one first RF coil, said first RF coil being sensitive to a magnetic field along a first axis;

at least one second RF coil, said second RF coil being sensitive to magnetic field along a second axis which is orthogonal to the first axis;

a signal combining circuit operatively connected to the first and second RF coils, said signal combining circuit having (1) the quadrature combining mode in which it quadrature combines signals received by the first and second RF coils and (2) the phased array mode in which it passes signals received by the first and second RF coils to corresponding receivers without combining said signals; and

a switch assembly connected to the signal combining circuit, said switch assembly switching the combining circuit between the quadrature combining mode and the phased array mode.

2. The magnetic resonance apparatus according to claim 1, wherein the signal combining circuit includes:

a phase shift circuit operatively connected to at least one of the first RF coil and the second RF coil for phase shifting received signals by 90.degree. relative to one another.

3. A magnetic resonance apparatus having a means for generating a main magnetic field through an examination region, a means for exciting magnetic resonance dipoles in the examination region, a multi-mode RF coil assembly which receives magnetic resonance signals from the resonating dipoles, and at least two receivers which receive and demodulate output signals from the multi-mode RF coil assembly, the multi-mode RF coil assembly comprising:

at least one first RF coil assembly, said first RF coil assembly being sensitive to a magnetic field along a first axis;

at least one second RF coil assembly, said second RF coil assembly being sensitive to magnetic field along a second axis which is orthogonal to the first axis;

a signal combining means operatively connected to and disposed on the first and second RF coil assemblies, said signal combining means having (1) a quadrature combining mode in which it quadrature combines signals received by the first and second RF coil assemblies and (2) a phased array mode in which it passes signals

received by the first and second RF coil assemblies to corresponding receivers without combining said signals; and

a switch means connected to the signal combining means and disposed adjacent the combining circuit on the RF coil assemblies, which switch means switches the combining means between the quadrature combining mode and the phased array mode without modifying the first and second RF coil assemblies, said switch means including a pair of diodes which are responsive to a DC biasing potential.

4. The magnetic resonance apparatus according to claim 3, wherein the presence of the DC biasing potential switches the receiver assembly to the quadrature combining mode.

5. The magnetic resonance apparatus according to claim 3, wherein the absence of the DC biasing potential switches the receiver assembly to the phased array mode, in the phased array mode, the at least two receivers receive an output signal sensed by the first RF coil and an output signal sensed by the second RF coil.

6. A magnetic resonance apparatus having a main magnet which generates a main magnetic field through an examination region, a radio frequency (RF) transmitter coil positioned about the examination region such that it excites magnetic resonance dipoles therein, an RF transmitter which drives the RF transmitter coil, a multi-mode RF receiver coil assembly which receives magnetic resonance signals from the resonating dipoles, and at least two receivers which receive and demodulate output signals from the multi-mode receiver coil assembly, the multi-mode receiver coil assembly comprising:

at least one first RF coil, said first RF coil being sensitive to a magnetic field along a first axis;

at least one second RF coil, said second RF coil being sensitive to magnetic field along a second axis which is orthogonal to the first axis;

a signal combining circuit operatively connected to the first and second RF coils, said signal combining circuit having:

(1) a phase shift circuit operatively connected to at least one of the first and second RF coils for phase shifting received signals by 90.degree. relative to one another;

(2) a quadrature combining mode in which it quadrature combines signals received by the first and second RF coils, in the quadrature combining mode, one of the at least two receivers receive a quadrature output signal and another of the receivers receives an anti-quadrature output signal from the signal combining circuit; and

(3) a phased array mode in which it passes signals received by the first and second RF coils to corresponding receivers without combining said signals; and

a switch assembly connected to the signal combining circuit, said switch assembly switching the combining circuit between the quadrature combining mode and the phased array mode.

7. The magnetic resonance apparatus according to claim 6, said apparatus further comprising:

a quadrature reconstruction processor which reconstructs the quadrature output signal into a quadrature image representation;

an anti-quadrature reconstruction processor which reconstructs the anti-quadrature output signal into an anti-quadrature image representation; and

an adder which adds the quadrature image representation and the anti-quadrature image representation.

8. A multi-mode magnetic resonance method including:

generating a main magnetic field through an examination region;

transmitting RF signals into the examination region to induce magnetic resonance in

nuclei;

receiving the induced magnetic resonance signals using a first pair of RF quadrature coils and a second pair of RF quadrature coils, said first and second pairs of RF coils being disposed in a linear array;

switching between one of a quadrature combination mode and a phased array mode;

in the quadrature combination mode, (i) phase shifting the received signals and (ii) combining the received, phase shifted signals into a quadrature signal and an anti-quadrature signal;

in the phased array mode, passing the received magnetic resonance signals uncombined;

demodulating the received magnetic resonance signals; and

reconstructing the demodulated signals into an image representation.

9. The method according to claim 8, wherein the selecting step includes:

optionally applying a DC bias to a switch assembly to select the quadrature combination mode.

10. The method according to claim 8, further including:

reconstructing the quadrature signal into a quadrature image representation;

reconstructing the anti-quadrature signal into an anti-quadrature image representation; and

combining the quadrature anti-quadrature image representations.

11. A multi-mode RF assembly for use in a magnetic resonance apparatus, the multi-mode RF assembly comprising:

a first RF coil assembly including at least one RF coil, said first RF coil assembly being sensitive to a magnetic field along a first axis to generate a first resonance signal;

a second RF coil assembly including at least one RF coil, said second RF coil assembly being sensitive to a magnetic field along a second axis which is orthogonal to the first axis to generate a second resonance signal 90.degree. out of phase from the first resonance signal;

a signal combining circuit which phase shifts and additively combines the phase shifted first and second is resonance signals into a quadrature output and subtractively combines the phase shifted resonance signals into an anti-quadrature output; and

a switch assembly which switches between outputting one of (i) a combined quadrature signal and a combined anti-quadrature signal and (ii) the uncombined first and second resonance signals.

12. A method of quadrature operation in a magnetic resonance apparatus, the method including:

(a) generating a temporally constant magnetic field through an examination region;

(b) transmitting RF signals into the examination region to induce magnetic resonance in nuclei;

(c) detecting induced magnetic resonance signals in quadrature using a quadrature coil assembly;

(d) combining and phase shifting by 90.degree. the detected signals into a quadrature signal and an anti-quadrature signal using a quadrature adder;

(e) transferring the quadrature and anti-quadrature signals to a pair of receivers;

and

(f) reconstructing the received quadrature and anti-quadrature signals into an image representation.

13. The method according to claim 12, wherein step (f) includes:

reconstructing the quadrature signal into a quadrature image representation;

reconstructing the anti-quadrature signal into an anti-quadrature image representation; and

combining the quadrature image representation and the anti-quadrature image representation.

DOCUMENT-IDENTIFIER: US 6150816 A
TITLE: Radio-frequency coil array for resonance analysis

Abstract Text (1):
An RF coil array which includes first and second RF coils that are overlapped to eliminate their coupling (to maintain zero mutual inductance between them) through space, a third coil connecting the first and second coils such that there is no net coupling between the first two coils through the third coil, and in which all three coils are well isolated from one another at the resonance frequency or frequencies of interest.

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Brief Summary Text (2):
The present invention relates to resonance systems, such as magnetic resonance imaging (MRI), nuclear quadrupole resonance (NQR), electron spin resonance (ESR) systems, and more particularly to a radio-frequency (RF) coil array and method for use in such systems.

Brief Summary Text (4):
NMR or MRI

Brief Summary Text (5):
In MRI systems or nuclear magnetic resonance (NMR) systems, a static magnetic field B.sub.o is applied to the body under investigation to define an equilibrium axis of magnetic alignment in the region of the body under examination. An RF field is then applied in the region being examined in a direction orthogonal to the static field direction, to excite magnetic resonance in the region, and resulting RF signals are detected and processed. Generally, the resulting RF signals are detected by RF coil arrangements placed close to the body. See for example, U.S. Pat. No. 4,411,270 to Damadian and U.S. Pat. No. 4,793,356 to Misic et al. Typically, such RF coils are either surface type coils or volume type coils, depending on the particular application. Normally separate RF coils are used for excitation and detection, but the same coil or array of coils may be used for both purposes. For multiple surface RF coils for use in NMR, see U.S. Pat. No. 4,825,162 to Roemer, et al.

Brief Summary Text (6):
A further increase in S/N can be realized with the use of quadrature coils as compared to the conventional linear coil designs. See for example U.S. Pat. No. 4,467,282 to Siebold and U.S. Pat. No. 4,707,664 to Fehn. Also see U.S. Pat. Nos. 4,783,641 and 4,692,705 to Hayes for a quadrature volume coil, commonly referred to as the "birdcage" coil in the NMR community. For the use of multiple volume coils for use in NMR, see U.S. Pat. No. 5,258,717 to Misic, et al., and the reference article by Leussler for the use of multiple volume coils for simultaneous head and neck imaging (See, C. Leussler, "Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck", SMRM 12th Annual Meeting, New York, Book of Abstracts, page 1349, 1993). Also, reference is made to commonly assigned U.S. patent application Ser. No. 08/745,893 filed on Nov. 8, 1996 titled "Radio-Frequency Coil and Method for Resonance Imaging/Analysis", and Ser. No. 08/993,932 entitled "Improved Radio-Frequency Coil and Method for Resonance Imaging/Analysis", filed on Dec. 18, 1997, the disclosures of which are incorporated herein by reference, for the use of

multiple volume and surface coils for use in NMR imaging.

Brief Summary Text (7):
The recent introduction of array coils to NMR, has led to commercially available cervical-thoracic-lumbar (CTL) array coil for entire spine imaging, and flexible body arrays for torso imaging. These multichannel coils significantly help reduce scan times. A routine MR study takes approximately 45 minutes, including the patient placement. This is uncomfortable especially for claustrophobic patients in general. In addition, prolonged scan times make the contrast-enhanced studies even more difficult to obtain. The almost 1 hour MR study with and without the contrast agent makes MR not so suitable for imaging emergency trauma cases.

Brief Summary Text (8):
This necessitates a new array coil with high S/N, that will allow the MR study of the torso, head, spine or joints such as the knee, wrist and shoulder etc., to be performed in reduced scan times. This will significantly reduce patient discomfort and increase patient throughput in a MR scanner. The reduced scan times will also allow MRI systems to be used in scanning emergency trauma patients.

Brief Summary Text (9):
A new area of MRI namely functional MRI or more commonly referred to as fMRI has emerged in the recent years. This technique provides the capability of mapping the brain functions, non-invasively using MR. Unfortunately, a major drawback of this technique is its lack of sensitivity. Once again, a coil with improved S/N will provide a much clear image that will assist in the diagnosis of disorders in the human brain.

Brief Summary Text (11):
NQR is a technique that is capable of locating and uniquely identifying nitrogen for the detection of explosives and/or narcotics, even when contained and concealed by other materials. NQR has potential application in general and medical imaging and industrial measurements, in addition to the detection of either explosives (including land mines) or narcotics. See U.S. Pat. Nos. 5,594,338 and 5,592,083 for the design of an RF coil employed in the NQR system.

Brief Summary Text (12):
Generally, a significant factor in contraband detection by means of NQR is that quadrapolar nuclei that are commonly present, and potentially readily observable, in narcotics and explosives include nitrogen (.sup.14 N) and chlorine (.sup.35 Cl and .sup.37 Cl), among possible other nuclei. Thus, in commercial applications it is necessary to be able to detect quadrapolar nuclei contained within articles of mail, mail bags or airline baggage, including carry-on and checked luggage. The resonant frequencies of the nitrogen and chlorine in these substances are different for each chemical structure, but are well defined and remain consistent. That is, for a given chemical structure the resonance frequencies for nitrogen and chlorine compounds remain intact and do not change, unless their chemical structure is altered.

Brief Summary Text (13):
Generally, NQR frequencies of quadrapole nuclei lie within 0.5-5 MHz range. However, for organic chlorine compounds, .sup.35 Cl chemical shifts range from 16-55 MHz. The chemical shift of chlorinated hydrocarbons occurs between 32-45 MHz. This is a very wide frequency range for one single turn RF coil of the aforementioned '338 and '083 references to cover.

Brief Summary Text (14):
Even the 0.5 to 5 MHz (a ten fold frequency) range of detection for .sup.14 N in explosives and or narcotics mandate a capacitance of a factor of 100 (f.sup.2 .varies.1/LC) to tune the coil from 5 to 0.5 MHz range, which are overwhelmingly large range of capacitances required to tune the RF coil. Since, the same RF coil was used for a wide frequency range, the RF coil design was un-optimized for the several frequency ranges of operation. This may affect the performance of the RF coil (Q values) and the entire NQR system (transmitter power, S/N), in the detection of low levels of nitrogen compounds found in plastic explosives and narcotics.

Brief Summary Text (15):
This necessitates that the RF coil design be optimized for maximum S/N over at least a majority of the frequency ranges of NQR operation and detection in reduced examination times.

Brief Summary Text (17):
Birdcage Coil

Brief Summary Text (18):
Even after several years following the introduction of array coils to NMR, the only coil that is commercially used for scanning the human head in a horizontally oriented B.sub.o magnetic field is the quadrature birdcage coil of Hayes '705. Other applications of this coil design are for the whole body, knee and wrist imaging. A birdcage coil consists of two rings connected by several straight segments referred to as legs. A planar schematic of an eight leg high-pass birdcage is shown in FIG. 1a. This coil consists of two end rings R1 and R2 and 8 legs 1 through 8. Each ring section between two legs are interrupted by two series 2C value capacitors. Their combined effect is one capacitor of C value. FIG. 1b is the front view of the birdcage describing the location of the ring with respect to the legs and includes the mode orientation. FIG. 1c is the side view of the coil outline shown for brevity.

Brief Summary Text (19):
The birdcage which is of the distributed inductance-capacitance type structure has several frequency modes. Of interest is the first or principal or k=1 quadrature mode. This k=1 quadrature mode has two linear components (1a, 1b), oriented orthogonal to one another as shown in FIG. 1b. As mentioned above, the quadrature coil provided a 41% improvement over the conventional linear coil designs. The birdcage expended half the power when compared to the conventional linear coil, thus significantly reduced the RF power deposited in the patient. The higher order or k>1 modes had no net field at the coil center and generally were not used for imaging. At the k=1 mode, the currents in the coil were cosinusoidally distributed such that the resultant field displayed a homogeneous distribution over the imaging field-of-view (FOV). It is for these regions this coil gained popularity in the NMR community for the several volumetric applications (torso, head, knee, wrist, etc.).

Brief Summary Text (20):
The dashed lines of FIGS. 1a, 1b and 1c are planes of symmetry for this birdcage. From FIG. 1b, there are four such planes (I, II, III, IV), that are distributed azimuthally (due to symmetry). There is one additional axial plane (V) that is centrally located between the two end rings R1 and R2, dissecting the coil axis (see FIG. 1c) which, in addition is also a virtual ground plane. The points where the planes of symmetry intersect the birdcage are "a, b, c, d, e, f, g, h" on ring R1 and "i, j, k, I, m, n, o, p" on ring R2, "q, r, s, t, u, v, w, x" on legs 1, 2, 3 . . . 8 respectively of FIG. 1a. Since points "q-x" are located on the virtual ground plane, these points are at virtual ground potential or have no net potential.

Brief Summary Text (21):
Should points "a-p" on the end rings be connected as shown in FIG. 1d, then the 8 leg coil of FIG. 1a will become a 16 leg coil of FIG. 1d and the frequency mode structure including the current distribution will be altered. The resultant structure in this case was still a single birdcage, even after the addition of eight more legs. Thus the increase in S/N was not realized even after this addition, although the homogeneity along the axial planes of the coil may have improved slightly over the eight leg coil. And since no increase in S/N was realized, this approach was unacceptable.

Brief Summary Text (22):
However, should the virtual ground points "q-x" in the legs of FIG. 1a be shorted, this will result in the coil of FIG. 1e. This will give rise to a new RF gradient mode, bi-phasic in nature with + & - lobes along the coil axis. However, it is noted that a RF gradient mode for the coil of FIG. 1e, has no net field at the coil center (i.e., the RF gradient mode has no net field in the central virtual ground plane of FIG. 1c). Therefore, although FIG. 1e has two birdcages that share one end ring R.sub.12 and even a new mode is realized, no net increase in S/N at the coil center is realized.

Brief Summary Text (23):
3-Channel Distributed Type Coil Head Array

Brief Summary Text (24):
A quadrature, 3-channel head coil was described by the inventor in previously filed Ser. No. 08/993,932, which provided improved S/N at the coil center and toward the top of the head (see FIG. 2). The coil consisted of two birdcages (coils #1, #2),

one distributed, quadrature modified surface coil (coil #3) and passive circuits were used for decoupling individual coils and to minimize the cross-talk between all coils in the array. The coil was operated in the multiple operating modes, with focus to the upper or lower portions of the brain or for routine head studies in one clinical setting, with high S/N and without compromising homogeneity. Here the birdcage, coil #2 and the quadrature surface coil #3 were asymmetrically overlapped and therefore isolated from one another and is the subject of previously filed U.S. Ser. No. 08/745,893.

Brief Summary Text (25):
The combination of coils #2 and #3 was then overlapped with birdcage coil #1. Since all three coils in the array were physically separated from one another, and were overlapped to maintain minimal mutual coupling, each coil in the array maintained their own RF current distribution and mode orientation. Several passive coil-to-coil decoupling electronics helped minimize the residual cross-talk between coils in the array. Each quadrature coil signal was routed to individual low noise figure, high gain preamplifiers before digitization. Diode protection circuits were inserted between the coil and the respective preamplifier for preamplifier protection during whole body transmit.

Brief Summary Text (26):
2-Channel Birdcage, Head and Neck Array

Brief Summary Text (27):
A quadrature, 2-channel birdcage array was described in C. Leussler, "Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck", SMRM 12th Annual Meeting, New York, Book of Abstracts, page 1349, 1993 for simultaneous head and neck imaging (see FIG. 3). This coil involved 2 birdcages; a coil 10 for the head and a coil 12 for the neck. The head birdcage 10 had eight fold symmetry, whereas the neck birdcage 12 had only a four-fold symmetry. The neck birdcage 12 had shoulder cut-outs for accomodating the neck as shown in FIG. 3. This coil provided an extended FOV without significantly compromising S/N and homogeneity over the extended FOV. Nevertheless, no increase in S/N was realized over extended FOVs. That is, the S/N of the array coil was comparable to individual head or neck coils over the head & neck scan volume.

Brief Summary Text (29):
The distributed surface coil of FIG. 4a has three meshes, 4 vertical segments referred to as legs and 2 horizontal segments referred to as ring segments. Each of the ring segments between the legs are broken with two 2C capacitors in series. Like wise, each of the end ring segments are also populated with two 2C value capacitors. For details of this coil design, refer to U.S. Pat. No. 4,783,641 to Hayes, et al.

Brief Summary Text (32):
A multiple surface coil arrangement disclosed in U.S. Pat. No. 5,256,971 to Boskamp is shown in FIG. 4d. Here, two surface coils of similar dimension are overlapped for minimum mutual inductance from one another. A third coil is added to this set, such that the third coil is magnetically isolated from the first and second coils. Here, all three coils are mutually isolated from another. In doing so, the third coil has a different coil geometry than the first and second coils, and extends beyond the FOV of the first and second coils combined.

Brief Summary Text (34):
This necessitates a coil system where the individual coils in the system are well isolated from one another and still maintain its current distribution and preferred mode orientation irrespective of its shape.

Brief Summary Text (36):
Solenoid Coil for NMR

Brief Summary Text (37):
One of the oldest and perhaps the most popular coil design that is commercially utilized for the several volumetric applications (torso, head, spine, knee, wrist) is of the solenoid design. See for example, U.S. Pat. No. 4,398,148 to Barjhoux et al.

Brief Summary Text (38):
FIG. 5a is one example of a solenoid head coil configuration commonly used in the NMR community. The N-turn solenoid is resonated with two series connected 2C value

capacitors. This coil has 2 planes of symmetry, I and II, respectively. Plane I intersects the coil at virtual ground points "a , b". A side view of the coil outline along with the head and the central virtual ground plane I is shown in FIG. 5b, for brevity.

Brief Summary Text (39):
Shorting the two virtual ground points of FIG. 5a will result in FIG. 5c. This will give rise to a new RF gradient mode along the coil axis. It will be noted that a RF gradient mode has no net field at the coil center (i.e. the RF gradient mode has no net field in the central virtual ground plane of FIG. 5b). Therefore, although FIG. 4c has two solenoid coils sharing the two virtual ground point "a, b" of FIG. 5a and even a new gradient mode is realized, the homogeneous mode of FIG. 5a will not be affected and no net increase in S/N is realized at the coil center.

Brief Summary Text (41):
A single turn solenoid coil of FIG. 6 was used to detect the .sup.14 N signals in crystalline form for detecting concealed explosives and narcotics employing nuclear quadrupole resonance (NQR). See U.S. Pat. Nos. 5,594,338 and 5,592,083 for the design of an RF coil employed in the NQR system.

Brief Summary Text (42):
FIG. 6 has one single turn RF coil which is tuned to a wide range (approx 0.5 to 5 MHz), by simply adding large and small value capacitances for coarse and fine tuning with the help of relay switches. As seen, the upper frequency range was ten fold of the lower range which mandated a 100 fold change in capacitance to tuned the coil. Since, the same RF coil was used for a wide frequency range, the RF coil design was un-optimized for the several frequency ranges of operation. This may affect the performance (transmitter power, S/N) of the RF coil and the entire NQR system, in the detection of low levels of nitrogen and chlorine compounds found in plastic explosives and narcotics.

Brief Summary Text (43):
This necessitates that the RF coil design be optimized for at least a majority of the frequency ranges of NQR operation and detection which will also help in reducing examination times.

Brief Summary Text (44):
This RF coil design will allow for at least one optimized coil in the array that will cover a part of the frequency spectrum, such that all coils in the array combined cover the entire frequency spectrum required for detection. This will help reduce the overall scan frequency range per coil and thus allow rapid tuning of coils in the array. This RF coil design may also be designed to allow for multiple tuning of the coils in the array without crosstalk and capable of simultaneous operation, which will help scan the entire frequency range in reduced scan times.

Brief Summary Text (45):
It is therefore a primary objective of the present invention to further improve S/N and reduce scan times of all such coil systems used for resonance imaging or spectroscopic analysis mentioned above. Specific applications of the coil described herein in accordance with the present invention include distributed type surface and volume coils, and single and multiple turn solenoid type coils.

Brief Summary Text (47):
The present invention provides an RF coil with high signal-to-noise ratio (S/N) over the imaging or spectroscopic field-of-view (FOV). The RF coil of the present invention enables one to reduce scan times and therefore patient discomfort without significantly compromising image quality. The RF is capable of operating in different FOVs in the multiple operating modes in one or multiple frequencies. Furthermore, the present invention provides a coil array capable of simultaneous operation in at least one frequency range.

Brief Summary Text (48):
A primary objective of the present invention is to provide a novel RF coil design with high S/N, capable of array operation in the single or multiple frequencies. Another objective is to provide an array design, which will provide a high combined S/N than any one coil operated alone. Yet another objective is to provide a RF coil capable of simultaneous multiple frequency operation for resonance imaging/spectroscopic analysis. A further objective is to have coils in the array that are well isolated from one another and maintain their individual current

distributions and mode orientations irrespective of the shape of the coil.

Brief Summary Text (49):
The design of the inventive coil involves first and second RF coils, that are overlapped to eliminate their magnetic coupling (to maintain zero mutual inductance between them) through space, a third coil physically connecting the first and second coils such that there is no net coupling between the first two coils through the third coil, and all three coils are well isolated from one another at the resonance frequency or frequencies of interest.

Brief Summary Text (50):
Please note all three coils in this coil system (first+second+third), may be volume type coils or surface type coils or a combination of both. A novel aspect of this invention is the unusual combination of coils in one integrated structure, that are well isolated from another and maintain their preferred current distributions and mode orientations.

Brief Summary Text (51):
In the embodiments of the present invention, the third coil has a FOV nearly identical to that of the combined FOV of the first two coils, and overlaps the combined FOV of the first two coils, such that, the S/N of all three coils combined is greater than any one coil operated alone. In other embodiments, several (first+second+third=integrated) such integrated coils are overlapped for minimal mutual inductance and used in an array configuration. The three individual coils in any one integrated design may be tuned to one or more resonance frequencies, for simultaneous use in imaging or spectroscopic analysis. Depending on the imaging FOV, individual coils in the array can be turned OFF or ON by the programmable transmit/receive (T/R) driver in the resonance system.

Brief Summary Text (52):
According to one particular aspect of the invention, a radio-frequency (RF) coil array for resonance imaging/analysis is provided. The coil array includes a first RF coil sensitive to RF signals produced during resonance imaging/analysis; a second RF coil located relative to the first RF coil with substantially zero coupling therebetween at a frequency or frequencies of the RF signals; and a third RF coil located relative to the first RF coil and the second RF coil such that there is substantially zero net current flow between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals.

Drawing Description Text (2):
FIG. 1a is a planar schematic view of a high-pass birdcage coil;

Drawing Description Text (3):
FIG. 1b is a is a schematic end view of the birdcage coil of FIG. 1a;

Drawing Description Text (4):
FIG. 1c is a side schematic view of the birdcage coil of FIG. 1a illustrating the plane of symmetry;

Drawing Description Text (5):
FIG. 1d is a planar schematic of a modified birdcage coil of FIG. 1a;

Drawing Description Text (6):
FIG. 1e is a planar schematic of a modified birdcage coil of FIG. 1a;

Drawing Description Text (7):
FIG. 2 is a schematic view of an RF coil having two birdcage coils, and modified spoke type quadrature surface coil;

Drawing Description Text (8):
FIG. 3 is a perspective view of a quadrature, 2-channel birdcage array;

Drawing Description Text (12):
FIG. 5a represents schematically a solenoid head coil;

Drawing Description Text (21):
FIG. 7f is a table illustrating various combinations of individual coils for

different modes of operation in accordance with the present invention;

Drawing Description Text (22):
FIG. 7g is a schematic illustration of a plurality of coils in combination in accordance with the present invention;

Drawing Description Text (26):
FIG. 8d is a schematic view of a 4-channel, quadrature head array in accordance with the present invention;

Drawing Description Text (27):
FIGS. 9a and 9b are front and side views of a 3-channel, quadrature knee array embodiment in accordance with the present invention;

Drawing Description Text (29):
FIGS. 10a and 10b are, respectively, schematic front and side views of a 3-channel, quadrature wrist array in accordance with the present invention;

Drawing Description Text (30):
FIG. 11a is a planar schematic view of a 3-channel, quadrature head and neck array in accordance with the present invention;

Drawing Description Text (33):
FIG. 12a is a schematic view of a distributed coil array for spine or torso imaging in accordance with the present invention;

Detailed Description Text (3):
Referring initially to FIG. 7a, the invention includes first (coil #1) and second (coil #2) RF coils, that are overlapped to isolate the coils from each other, by causing the net shared flux between the coils to be zero. A third RF coil (coil #3) of FIG. 7a, is superimposed on the combination of coils #1 and #2, and physically connects coils #1 and #2 at several points along the coil periphery (see FIG. 7b). For the sake of explanation, the coil #3 may connect to coils #1 & #2 at points A, B and A', B' respectively. This however is done such that there is no net coupling between coils #1 and #2 through coil #3. Thus coils #1 and #2 are isolated from one another and still maintain their individual current distributions and B field orientations. Also, the currents in coil #3 are undisturbed and maintain their original distribution and B field orientation. Thus, all three coils are well isolated from one another and perform the intended function in a resonance experiment independent of the other.

Detailed Description Text (4):
Only coil outlines are shown in FIG. 7a for brevity. Not shown are impedances (inductances & capacitances) needed to resonate the RF coil at the frequencies of interest. It will be appreciated that the individual coils of FIGS. 7a may be of the volume type or the surface type or their combination as is discussed more fully below in connection with the specific embodiments.

Detailed Description Text (5):
Individual current distributions of coils #1, #2 and #3 are shown in FIG. 7a. Accordingly, their resultant B field orientations are directed in to the plane of the paper (if the fingers of the right hand are curled in the direction of the current, according to the right hand rule, the resultant B field direction of the coil is in the direction of the thumb, and in this case will be pointing in to the plane of the paper). By way of overlapping coils #1 and #2 are isolated from one another and maintain their individual current directions and preferred mode orientations. Here mode is referred to as the frequency mode of interest for a resonance experiment. For the cases of NMR or NQR, the modes of interest may be for one or more distinct radio-frequencies. However, mode orientations are the orientations of the B field over the imaging or the spectroscopic field-of-view (FOV) for individual RF coils, at the frequencies of interest. That is, for a linear RF coil case, there exists one mode that is of interest and one mode orientation. However, for a quadrature RF coil case, there exists two linear modes that are oriented orthogonal to one another. These two linear modes however may be tuned to the same frequency resulting in a quadrature coil, or may be tuned to two distinct frequencies thus depicting a dual tuned RF coil with linear operation at both frequencies.

Detailed Description Text (6):

Although it is preferred that coils #1 and #2 maintain identical coil dimensions, it is not an absolute necessity. In fact included in this disclosure is a head and neck array, where coils #1 and #2 are not identical in dimension. Nevertheless for the sake of simplicity, coils #1 and #2 of FIG. 7b have identical dimensions. Coil #3 connected to this combination of coils #1 and #2, encompasses a larger FOV covered by coil #1 or #2 alone. In fact, the FOV of coil #3 in this case is not only comparable but also superimposes the combined FOV of coils #1 and #2.

Detailed Description Text (7):
The combined S/N of coils #1 and #2, at the coil center (at the region of overlap) may be close or equal to the S/N of coil #3 of FIG. 7a. Thus the combined S/N of all three coils, that are isolated from another will be substantially greater than any one coil in the array. This is because the direction of currents in all three coils will remain the same and have similar B field orientations. Hence signals from all three coils add up. Since they are isolated from one another the noises from the coils in the array are uncorrelated, resulting in a substantial increase in combined S/N. For details of the mathematical expressions of combined S/N, refer to equations 19 and 20 of U.S. Pat. No. 4,825,162 of Roemer et al.

Detailed Description Text (9):
Since the inventive design has three coils in one integrated system, all coils must be isolated from one another to reduce their cross-talk which is necessary to increase the combined S/N of an experiment. In order to cancel the magnetic coupling between neighboring coils, they must be overlapped to cancel their net shared flux. The flow chart of FIG. 7c, represents a procedure suitable for optimizing all coils in the integrated coil system of FIG. 7b.

Detailed Description Text (10):
In steps S1 and S2, the first coil #1 is built and tested individually. Next, in step S3 the second coil #2 is built. In step S4, coil #1 and #2 are overlapped to cancel their coupling. Namely, in step S5 it is determined whether coils #1 and #2 are isolated by a predefined acceptable amount (e.g., coupled by less than -20 dB). If no in step S5, the coils #1 and #2 are repositioned relative to each other in an effort to improve the isolation therebetween. Steps S4 and S5 can then be carried out until acceptable isolation is achieved. If yes in step S5, the combined coils #1 and #2 may be tested as represented in step S6.

Detailed Description Text (11):
Then coil #3 will be built and added to this assembly as represented in step S7. After this addition, should the isolation between coils #1 & #2 deteriorate as determined in step S8, then either coils #1 & #2 be overlapped to compensate for the cross-talk introduced by the addition of coil #3 or the mechanism of FIG. 7d be used or a combination of both can be used to reisolate coils #1 & #2 after the addition of coil #3. (Step S9). Overlapping coils #1 & #2 again will cancel the net mutual flux shared by coils #1 & #2 after the introduction of coil #3. Final testing of the assembled array can then be carried out in step S10 upon achieving acceptable isolation between the respective coils. Once this optimum overlap is determined, a relatively high precision of duplication can be achieved from one coil batch to another in mass production, by etching the two coils on one or both sides of a single printed circuit board. However in addition to the above, any cross-talk by way of current flow between coils #1 & #2 via coil #3 can be minimized or eliminated in some instances by following the mechanism of FIG. 7d. Furthermore, if there exists any residual cross-talk, this too can be minimized or eliminated by introducing electrical coupling cancellation networks, one example may be similar to that of U.S. Pat. No. 4,769,605 to Fox.

Detailed Description Text (13):
From FIG. 7b, coil #3 connects to coils #1 & #2 at points A, B & A', B', respectively. If points A, B & A', B' were mid-points between two identical capacitors, then their voltage will be an average of the two potentials spanned by the two identical capacitors. This average potential may be denoted as being equal to V. In the present case, when coils #1 & #2 are identical, points A & A' in coils #1 & #2 will be at equi-potential. Similarly, points B & B' will also be at equi-potential. Please note, points B & B' may or may not be at the same potential as points A & A', which will depend on either the coil symmetry, or the distrubution of impedances within the coil or a combination of both. Since there will be no current flow between points of equi-potential, there is not net coupling between coils #1 & #2 via coil #3 with its addition. That is, the isolation between coils #1 & #2 will remain virtually the same with or without the addition of coil #3. Note,

this will be true only if the net flux shared by coils #1 & #2 are close to zero or the cross-talk between them is almost negligible or both the above conditions are satisfied, before the adddition of coil #3.

Detailed Description Text (16):
It is by these ways (cancelling net mutual shared flux or isolating with any other scheme, proposed additional isolation scheme of FIG. 7d and/or that of Fox, with the above) the isolation between the above coils can be maintained if all coils were fixed or etched on a rigid or on a flexible printed circuit board. Note, an isolation of -20 dB was set as a target. In actuality, this value can be set to any other number based on the coil design and expected combined S/N. We set a -20 dB value, as this will relate to a 1% loss in combined S/N from the optimum value obtained at coil overlap for coils of identical dimension. For details of coil isolation values and its relation to S/N, please refer to the article by Tropp et al., in the Review of Scientific Instrumentation, Volume 62, Number 11, November 1991.

Detailed Description Text (17):
Finally, although it is advantageous to test the individual coils separately as they are built (like that shown in FIG. 7c), once the optimum settings of the coils and isolation values are engineered and specifications met, then simply final testing of the entire RF coil system (consisting of coils #1, #2 and #3) is advised. This will considerably cut the time and costs incurred in the final production line prior to product shipment. However, the proposed flow-chart is a methodical progression of the coil design which is also designed so to enable a relatively easier debugging or trouble shooting of the coil system when needed.

Detailed Description Text (18):
Individual Coil Coupling

Detailed Description Text (19):
A preferred method of coupling to individual coils in the RF coil array and interfacing to the system is shown in FIG. 7e. Let us assume the case when all coils in the array of FIG. 7b are in quadrature. Then there will be a total of six linear modes, operating at the resonance frequency of interest. These modes are a1 & a2 of coil #1, b1 & b2 of coil #2 and c1 & c2 of coil #3, respectively.

Detailed Description Text (20):
Generally, the linear modes of a coil are matched to 50 ohms using balanced matched capacitors (not shown) and connected to quadrature hybrids via baluns. Either 50 ohm "criss-cross" discrete network (X.sub.L =X.sub.C =50 ohms; L=124nH, C=50 pF for approx. 64 MHz or 1 .5 T) or shielded, transmission line networks such as coaxial cable traps tuned to frequencies very close to the resonance frequency may be used as baluns to convert the balanced feed to an unbalanced line (see FIG. 7e). This is done to isolate the coil grounds from the system ground and to prevent leakage of the circulating RF currents on the ground shield of the coaxial cable exitting the system.

Detailed Description Text (21):
These networks (discrete or transmission line) are shielded as shown by the dotted lines to minimize their interaction with the whole body transmit field and their interaction with the RF coils themselves. Note, it is not necessary to shield the discrete networks but are shown as the preferred embodiment. However, we prefer that the cable traps be shielded so the fields generated by the traps are contained to within the volume encompassed by their shield. This is also done so the cable trap can be made uni or bi directional, depending the nature of the trap's use. The cable trap shown can be made uni directional by shorting its shield to one side of the coaxial cable shield. Likewise it can be made bi-directional by floating the shield, as shown in FIG. 7e.

Detailed Description Text (22):
The cable trap consists of two turns 1" in diameter, is wound on a delrin spindle with groves, using a semi-rigid cable of 0.085" o.d. along with fixed and variable capacitors for tuning a specified frequency range centered around the resonance frequency of operation. The RF shield is approximately 1.5.times.1.5.times.0.5" in dimension.With inductor Q values (of that of L in the discrete or that created by the coaxial cable in the transmission line network) of approximately 175-200, impedances (with zero reactances=resistance) of approximately 8-12K.OMEGA. can be realized across the baluns, which is adequate to isolate the grounds at 64 MHz

(Resistance R=Q.omega.L). Thus coil to ground leakage and coil to coil interactions be minimized or eliminated and high RF coil efficiences can be maintained.

Detailed Description Text (23):
Then the linear signals are combined using a phase shifting network to create a single quadrature output per coil. This is followed by a diode protection network before the preamplifier. All three coils are actively decoupled during whole body transmit (circuitry not shown). Although this decoupling will achieve a-25 dB isolation per coil at the resonance frequency of operation, the additional series-shunt pin-diode protection circuit shown will provide a further -45 dB of isolation between the coil and the preamplifier in every channel. During whole body transmit, diode D1 is turned ON which shunts all the RF present in the signal line to ground before reaching the preamp. Diode D2 is reverse biased during transmit and helps further isolate the RF present in the signal conductor to the preamp input. During receive D1 is reverse biased and D1 is forward biased to allow all of the RF signals to the preamplifier before digitization and further amplification at the system receiver. Thus the diode circuit will ensure a safe preamp operation (preamp input maximum of roughly +20 dBm).

Detailed Description Text (24):
Thus all 3 coils are interfaced to 3 channels in a resonance receiving system. It is to be noted, that this way of interfacing the coils to the system is preferred. However, the outputs from the individual coils can be further combined and interfaced to fewer channels of the resonance system. For example, the quadrature outputs from coils #1 & #2 or from all coils #1, #2 & #3 can be combined prior to interfacing to the system, the latter case resulting eventually in a single channel. Likewise, individual modes from all coils can be interfaced to their respective channels (in this case six channels, two from each coil) of the transceiver of a resonance system.

Detailed Description Text (26):
Regardless of the frequency of operation, please note the individual coils of FIG. 7b can be turned ON or OFF using the programmable T/R drivers of the resonance system which will result in a total of 7 modes of operation for this 3 coil network, as shown in FIG. 7f.

Detailed Description Text (27):
For example in one configuration, coils #1 and #2 are turned ON and coil #3 is turned OFF. Likewise, coil #1 is turned ON and coils #2 and #3 are turned OFF. All such combination of coils are shown in FIG. 7f. Only two such combinations currently seem not possible, where coil #3 is ON and coil #1 is ON (coil #2 turned OFF) or where coil #3 is ON and coil #2 is ON (coil#1 turned OFF). This is because if one of two coils #1 or #2 are turned OFF, then coil #3 will couple to coils #2 and #1, respectively through space since the minimum mutual inductance condition was not achieved between them. However, it should be understood that one skilled in the art can counteract this unwanted coupling with the addition of electrical cancelling networks or some modification of the coils themselves or a combination of both.

Detailed Description Text (28):
RF Coil System Arrays

Detailed Description Text (29):
Finally, several of these integrated (coil #1+coil #2+coil #3) coil systems, may be overlapped to provide high combined S/N over extended FOVs, as shown in FIG. 7g.

Detailed Description Text (30):
In summary, coils #1, #2 and #3 may be linear or quadrature and of the volume type or surface type or their combination. Should the coils be of the volume type, the dashed line of FIG. 7b will be along the common coil axis. Also, the coils may be tuned to the same or different resonance frequencies. For example, coils #1, #2 and #3 may be tuned to the same resonance frequency. In an another example, coils #1 and #2 may be tuned to one resonance frequency, and coil #3 is tuned to another resonance frequency. In yet another example, the individual coils in the array are tuned to different resonance frequencies and capable of simultaneous operation.

Detailed Description Text (31):
A few examples of the above concept are extended to distributed type volume coils like the birdcage, distributed surface coils and solenoid type coils. Their specific applications to NMR and NQR are described as embodiments of the present invention

disclosure.

Detailed Description Text (32):
Embodiment #2--3 Channel, Quadrature Birdcage Array

Detailed Description Text (33):
Embodiment #2 of this invention is the receive only coil of FIG. 8a. This consists of two birdcages (#1, #2) overlapped to maintain minimum mutual inductance, such that the net flux shared by them is zero. Coil #1 consists of rings R1, R2 and eight legs (1,2,3 . . . 8) that connect them. This coil is resonated with C1 value capacitors. Coil #2 consists of R3 and R4 and eight legs that connect them and are also resonated with C1 value capacitors. Note, the 8 legs of coil #2 are co-linear to that of coil #1. Here, both coils #1 and #2 are identical in dimension. Coil #3 connects coil #1 and #2 at 16 points. Coil #3 comprises of rings R1 and R4 which are connected by eight legs (9, 10, . . . 16). Coil #3 is resonant with C1 and C2 value capacitors. All coils are tuned to the same resonance frequency of interest.

Detailed Description Text (34):
After the addition of C2 to coil #3, the isolation between coils #1 and #2 remained virtually the same, which means that the eight legs of coil #3 intercepted rings R1 and R4 at corresponding equi-potential points. In this particular case, the eight legs of coils #1 and #2 happened to be at the symmetry planes for coil #3. Thus there was no net coupling between coil #1 and #2 via coil #3. There was also no net coupling between coil #3 and coils #1 or #2. Thus each coils maintained their own current distribution and their own mode distributions.

Detailed Description Text (35):
FIG. 8b is a front view of the coil, and shows the location of the legs of the birdcage and their mode orientations of the six linear modes of coils #1, #2 & #3 (a1, a2, a3, b1, b2, b3). Closed dark dots are locations of the legs of coils #1 & #2 connecting end rings R1 & R3 to R2 & R4 respectively, whereas open circles are for coil #3 which connect to end rings R1 and R4 only.

Detailed Description Text (36):
FIG. 8c is a side view of the coil outlines, with a head cartoon. As seen, coil #1 coverage extends from the c2-c3 cervical-spine and extends to the top of the cerebellum, coil #2 coverage extends from the mid cerebellum to the top of the head, whereas coil #3 coverage spans the combined FOV's of coils #1 & #2, respectively. Thus, routine head scanning can be accomplished with enhanced S/N, which can be used to reduce scan time or enhance image resolution or a little bit of both can be accomplished with the inventive coil. Furthermore, where specific focus is needed, either coil #1 or coil #2 can be individually turned ON to scan different portions of the human brain. Coupling to the six linear modes of this coil and their interface to the system can be accomplished similar to FIG. 7e. Here, three quadrature coil outputs are interfaced to 3 channels of a NMR system.

Detailed Description Text (37):
Embodiment #3--4 Channel, Quadrature Head Array

Detailed Description Text (38):
A quadrature, 3 channel head coil was described by the author of this invention in the previously mentioned application Ser. No. 08/993,932. See FIG. 2 for details. The S/N of this prior art coil can be further improved over the entire brain with the addition of coil #3, of FIG. 8d. This is also similar to adding coil #4 in FIG. 8c, needed to provide improved S/N over regions in the top of the head.

Detailed Description Text (39):
Coils #1, #2 and #3 have eight legs. Coil #4 is of the self-shielded type and has a total of 16 legs (8 primary and 8 secondary). See the above-mentions application Ser. Nos. 08/745,893 and 08/993,932 for the details of the coil #4's design and construction. Each of the four quadrature outputs from the coils in the head array are interfaced to 4 channels of the resonance receiving system, in this case a NMR receiving system. Please note, all four coils have a shielded, tuned coaxial cable trap in addition to the coupling and interface electronics mentioned in FIG. 7e. These coaxial cable traps help further isolate the RF coil grounds at the preamplifier level to the system ground and interfaces the coil outputs to the system receiver.

Detailed Description Text (40):

Please note, individual coils in the array can be turned ON or OFF to image a smaller FOV than the entire coil. If the focus was on the upper parts of the brain, only coils #2 and #4 need be turned ON, whereas if the focus was on the mandible areas only coil #1 may be turned ON.

Detailed Description Text (41):
Embodiment #4--3 Channel, Quadrature Knee Array

Detailed Description Text (42):
FIG. 9a and 9b are front and side views showing the coil outlines of the knee array. Here, coils #1, #2 and #3 have 4 legs, each. Legs 1, 2, 3 and 4 belong to coils #1 and #2 while 5, 6, 7 and 8 belong to coil #3. All legs are azimuthally distributed as shown in FIG. 9a. Coils #1 and #2 are first overlapped to maintain minimum mutual inductance. Coil #3 is then added which physically connects to coils #1 and #2, such that there is no net coupling between coil #1 and #2 via coil #3. A side view of the coil outlines along with a knee cartoon is shown in FIG. 9b.

Detailed Description Text (43):
FIG. 9c is a modified knee array. Here, coils #1 and #2 have 8 legs (1,2,3 . . . 8) each, distributed in the fashion shown. These coils are first overlapped to maintain minimum mutual inductance. Coil #3 that physically connects coils #1 and #2 have only 4 legs (9,10,11,12) which are distributed symmetrically. This arrangement is done to image the foot and the ankle along in addition to imaging the knee and the human calf. Please note, the coils of FIGS. 9 may have a split-top to ease the patient access.

Detailed Description Text (44):
Embodiment #5--3 Channel, Quadrature Wrist Array

Detailed Description Text (45):
FIGS. 10a and 11b are front and side views of a 3 channel, quadrature wrist array. Coils #1 and #2 have 4 legs (1,2,3,4) and are overlapped for minimal mutual inductance. Coil #3 that connects coils #1 and #2 has four legs (5,6,7,8). Thus the entire wrist array has a total of 8 legs as shown in FIG. 10a. Please note, the opening of the wrist coil is elliptical in shape to accommodate imaging of the fingers of the human hand. This also facilitates lateral placement of the coil along side the patients body inside a MRI machine. This high S/N coil allows for high-resolution imaging of the carpal ligaments of the human wrist.

Detailed Description Text (46):
Embodiment #6--3 Channel, Quadrature Head and Neck Design

Detailed Description Text (47):
A planar schematic of the coil is shown in FIG. 11a. Coil #1 has 8 legs (1,2,3 . . . 8) and covers the head FOV. Coil #2 has 8 legs (1,2,3 . . . 8) and has shoulder cut outs to accomodate the entire human neck. Each of these coils are resonant at the NMR frequency. Coil #1 and #2 are overlapped for minimal mutual inductance. Coil #3 connects coil #1 and #2 at eight points and hence has 4 legs distributed at right angles from one another. Thus the entire head and neck coil has 12 legs. Here, coil #1 is resonant with C1, and coil #2 is resonant with C2 whereas coil #3 is resonant with C1, C2 and C3, respectively. A front view of the coil is shown in FIG. 11b.

Detailed Description Text (48):
Here, just coil #1 or coil #2 can be turned ON for performing head or neck only studies. Also, all coils (#1, #2 & #3) can be turned ON to perform extended FOV head and neck studies, simultaneously. In this case where coil #3 spans a similar FOV as the combined FOVs of coils #1 & #2, the signals add up and since the noises are uncorrelated, enhanced S/N will be realized, unlike the prior art of FIG. 3. Also, coil #4 of FIG. 8d, can be added to the 3 channel, quadrature coil of FIG. 11 to further improve the S/N toward the top of the head (see FIG. 11c).

Detailed Description Text (50):
FIG. 12a is a distributed surface coil array for brain of torso imaging. Here, coil #1 and #2 are overlapped to maintain minimum mutual inductance. Therefore, the net flux shared by these two coils is zero. Both these coils are identical in dimension. They comprise of 2 ring segments, 4 legs and are resonated with C1 value capacitors. These coils are matched to 50 ohms, across the terminals "a, b" similar to the circuit of FIG. 7e and interfaced to 2 channels of the MRI system. Evidently, these two outputs can be matched and summed using a phase shifter, resulting in a single

channel quadrature output.

Detailed Description Text (51):
Coil #3 consists of 2 ring segments and 3 legs that connect to coils #1 and #2. Coil #3 is tuned to the resonance frequency of interest with C1 and C2 value capacitors. Coil #3 is matched to 50 ohms across "c" terminal using the similar circuitry of FIG. 7e. After the addition of coil #3, the isolation between coils #1 and #2 remained virtually the same, is indicative of a well isolated system. Please note, one integrated RF coil unit I comprises of coils #1, #2 and #3, respectively.

Detailed Description Text (52):
FIG. 12b is an extension of FIG. 12a, where coil #3 has an additional ring segment resulting in coil #4 on the same coil system. This coil is tuned to the resonance frequency of interest with C1, C3 and C4. Coils #1 and #2 here are tuned with C1, whereas coil #3 is tuned with C1 and C3. The outputs "a, b" can either be routed to two receiver channels, or combined using a phase shifting network resulting in a single quadrature output. Similarly, the other two outputs, "c, d" can be either routed to two other receiver channels or combined prior to the receiver. Please note, one integrated RF coil unit I comprises of coils #1, #2, #3 and #4, respectively.

Detailed Description Text (53):
FIG. 12c is a result of many such integrated RF coil circuits I, II, . . . of FIGS. 12a or 12b, in an array configuration. The coils of FIGS. 12 may be used to image the spine or wrapped around the human torso for imaging the liver, kidney, heart, etc. They may also be used to scan both feet for imaging the blood flow.

Detailed Description Text (55):
The solenoid type volume coil of FIG. 13a has multiple uses. It can be used to image the human brain, knee, elbow, wrist, foot and ankle, torso in a vertical field NMR machine. This type of a coil may also be used in a NQR system and used to detect the explosives and narcotics in baggages, mails, etc. The NQR system may also be used as a security check at various public places, such as airports, railway stations,etc. and may be used to detect for plastic explosives, narcotics, etc.

Detailed Description Text (56):
FIG. 13a consists of a total of 3 solenoid coils. Coils #1 and #2 are identical in dimension. They both have 2 turns separated by a set distance and are tuned with C1 value capacitors. These coils are overlapped to maintain minimum mutual inductance, thus the net flux shared by these two coils are zero. Coil #3 is then introduced by shorting virtual ground points "a, b" in coil #1 to "c, d" in coil #2. However, the shorting between points "a, c" is done with two turns, the first turn exists in the virtual ground plane of coil #1 and the second turn in the virtual ground plane of coil #2. This is done such that coils #1 and #2 will not see coil #3. Also, this shorting is interrupted with C2 and the shorting between the points "b, d" is interrupted with C3 value capacitor. Please note, only two turns are used for coils #1 and #2, for simplicity. In practice, coils #1 and #2 may have N (N.gtoreq.1) turns, and coil#3 may have M (M=N or M.noteq.N) turns.

Detailed Description Text (57):
The resultant integrated structure I comprises of coils #1 and #2 that are overlapped for minimum mutual inductance and coil #3 physically connecting coils #1 and #2, such that there is no net coupling between coils #1 and #2 via coil #3. In this preferred case for human imaging, all coils are tuned to the same NMR frequency. However for the NQR case, all coils may be tuned to the same or different frequencies.

Detailed Description Text (58):
A preferred embodiment is where all coils are tuned to different NQR frequencies and have their own tunable range of frequencies, lets say for example coil #1 covers from 0.5-1.5 MHz, coil #2 from 1.5-3.0 MHz and coil #3 from 3-5 MHz, respectively. Each coil design is optimized to cover this frequency range and has its own capacitor bank as shown in FIG. 13b to tune the specified frequency range. The individual switches may be computer controlled (not shown) to tune the individual RF coil to the specified resonance frequencies. The object that need to be scanned is introduced along the coil axis.

Detailed Description Text (59):
Alternately, the solenoid design may be adapted to a surface type design and may be

used for surface detection of drugs, narcotics, explosives, etc. The RF coil may also be used in a quasi surface--volume type design as well, for several medical and non-medical applications.

Detailed Description Text (60):
FIG. 14 is a system block diagram, which illustrates the utility of the RF coil of the pressent invention in NMR imaging and spectroscopy, for example. The system has a main magnet which covers the time varying gradient coils, an RF shield that isolates the RF coil from the gradient coils and a whole-body RF coil most commonly used for uniform B field transmit over a large imaging FOV. The main magnet strength sets the NMR frequency of operation. The time varying gradient fields help spatially encode the NMR signals. The RF whole body coil is used to transmit, while the local RF coil is used to pick up the NMR signals from the object under investigation (NMR phantom). A number of receiver coils may be used in an array configuration and may be summed either analog or digitally to produce the resultant image. Signals from the several receiver ports may be acquired via one or multiple receiver channels. An n-to-1 channel multiplexer is shown in the drawing. This helps by-pass n channel coil data to use one channel of the NMR system. Alternatively, an n channel NMR system may also be used.

Detailed Description Text (61):
Similarly, a NQR system may be realized but without the main magnet of FIG. 14. The resonance frequencies are set by the different nuclei themselves. Thus there is no need for the main magnet of FIG. 14. The time varying gradients of FIG. 14 may or may not be used, depending on the technique used to map the chemical species.

Detailed Description Text (62):
From all the above description, for someone skilled in the art, is must now be apparent that the inventive novel concept of FIG. 7 may be adapted to a number of different coil designs for the several resonance techniques, such as NMR, NQR, etc. It must also now be apparent that the individual coils in an integrated RF coil system may be tuned to the same or different frequencies.

Detailed Description Text (63):
It is to be noted that the individual coils in the array may be shaped in such a way to provide a high S/N and uniform coverage over the imaging FOV. The coils may be used to image in the different operating modes. The signal may be combined prior to the preamplifier or post the preamplifier in analog or digital fashion. The individual coils in the array may be tuned to one or more frequencies.

Detailed Description Text (64):
It must be further apparent that the coil designs in the distributed cases may be of the low-pass, high-pass, band-pass, band-stop or a combination of the above different configurations. Also, the coils may be of the volume type, surface type or a combination of both. Individual coils in the array may be linear or in quadrature. The coils may be used for transmit only, receive only or may be used for transmit and receive purposes. Individual coils in the array may be interfaced to separate channels in the multi-channel resonance system or may be time-multiplexed to one or more channels of a single or multi-channel resonance system.

Detailed Description Paragraph Table (1):
Embodiment # Type Description Application
1. Preferred embodiment described above 2. Volume Birdcage NMR - for brain imaging 3. Volume Birdcage NMR - for brain imaging 4. Volume Birdcage NMR - for knee imaging 5. Volume Birdcage NMR - for wrist imaging 6. Volume Birdcage NMR - for head and neck imaging 7. Surface Distributed NMR - for spine or torso imaging 8. Volume Solenoid NMR - for brain, knee, knee, elbow, wrist, foot or ankle, torso NQR - security checks for detection of narcotics, plastic explosives, etc.

Other Reference Publication (2):
Michael Burl, Ian R, Young, Examples of the Design of Screened and Shielded RF Receiver Coils, pp. 326-330.

Other Reference Publication (3):
Srinivasan, Improved Radio-Frequency Coil and M Ethod for Resonance/Imaging Analysis, U.S. Patent Application No. 08/993,932, filed Dec. 18, 1997.

Other Reference Publication (5):

"Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck" by C. Leussler SMR 1993.

Other Reference Publication (7):
"Examples of the Design of Screened and Shielded RF Receiver Coils"; Michael Burl and Ian R. Young, pp. 326-330.

CLAIMS:

1. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil electrically connected and located relative to the first RF coil and the second RF coil such that there is substantially zero net current flow between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals.

2. The coil array of claim 1, wherein the first RF coil, second RF coil and third RF coil are sufficiently isolated from one another to maintain predefined current distributions and mode orientations for the respective coils.

3. The coil array of claim 1, wherein the third RF coil has a field-of-view which is similar as compared to a combined field-of-view of the first RF coil and the second RF coil.

4. The coil array of claim 1, wherein each of the first RF coil, second RF coil and third RF coil are volume type coils.

5. The coil array of claim 4, wherein each of the first RF coil, second RF coil and third RF coil are birdcage type coils.

6. The coil array of claim 5, wherein the coil array is sized to receive at least one of a human head, a human knee, and a human wrist within the first RF coil second RF coil and third RF coil.

7. The coil array of claim 5, further comprising a fourth RF coil positioned toward an end of the coil array.

8. The coil array of claim 5, wherein the coil array is sized to receive a human head.

9. The coil array of claim 5, wherein the coil array is sized to receive a human head and neck.

10. The coil array of claim 4, wherein each of the first RF coil, second RF coil and third RF coil are solenoid type coils.

11. The coil array of claim 10, wherein the coil array is sized to receive at least one of a human head, knee, wrist or torso within the first RF coil, second RF coil and third RF coil.

12. The coil array of claim 1, wherein each of the first RF coil, second RF coil and third RF coil are surface type coils.

13. The coil array of claim 12, further comprising a fourth RF coil.

14. The coil array of claim 1, wherein the net shared magnetic flux between the first RF coil and the second RF coil is substantially zero.

15. The coil array of claim 1, wherein the third RF coil is physically connected to the first RF coil and the second RF coil.

16. The coil array of claim 1, wherein the first RF coil and the second RF coil maintain substantially similar physical dimensions.

17. The coil array of claim 1, wherein each of the first RF coil, second RF coil and third RF coil is configured to provide a quadrature output.

18. The coil array of claim 1, further comprising means for selectively turning the first RF coil, second RF coil and third RF coil on and off to control a mode of operation.

19. The coil array of claim 1, wherein at least one of the first RF coil, second RF coil and third RF coil is a volume type coil, and at least another of the first RF coil, second RF coil and third RF coil is a surface type coil.

20. The coil array of claim 1, wherein the first RF coil, second RF coil and third RF coil are tuned to the same resonance frequency.

21. The coil array of claim 1, wherein the first RF coil and the second RF coil are tuned to one resonance frequency, and the third RF coil is tuned to another resonance frequency.

22. The coil array of claim 1, wherein the first RF coil, second RF coil and third RF coil are tuned to respective different resonance frequencies.

23. A system comprising the coil array of claim 1, and further comprising means for driving the coil array during imaging/analysis.

24. A resonance imaging/analysis system, comprising:

an RF coil as recited in claim 1; and

means for processing RF signals which are at least one of received from the RF coil and transmitted from the RF coil in order to obtain a resonance image/analysis.

25. The coil array of claim 1, wherein the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

26. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil electrically connected and located relative to the first RF coil and the second RF coil such that there is substantially no net coupling between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals.

27. The coil array of claim 26, wherein the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

28. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil located relative to the first RF coil and the second RF coil such that each of the first RF coil, second RF coil and third RF coil are substantially isolated from the other coils at the frequency or frequencies of the RF signals;

wherein a field-of-view of the third RF coil substantially overlaps and is substantially similar to a combined field-of-view of the first and second RF coils.

29. The coil array of claim 28, wherein the third RF coil is electrically connected to the first RF coil and the second RF coil.

30. The coil array of claim 28, wherein

the third RF coil is electrically connected to the first and second RF coils, and

the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

31. The coil array of claim 30, wherein the third RF coil is electrically connected to the first RF coil and the second RF coil.

32. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil located relative to the first RF coil and the second RF coil such that there is substantially no net coupling between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals;

wherein a field-of-view of the third RF coil substantially overlaps and is substantially similar or larger than a combined field-of-view of the first and second RF coils.

33. The coil array of claim 32, wherein

the third RF coil is electrically connected to the first and second RF coils, and

the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

DOCUMENT-IDENTIFIER: US 5898306 A
**** See image for Certificate of Correction ****
TITLE: Single circuit ladder resonator quadrature surface RF coil

Abstract Text (1):
A single-circuit quadrature surface coil is formed from two ladder resonator coils and includes a first mode circuit path for detecting or generating magnetic flux in a vertical axis from a body under investigation and a second mode circuit path for detecting or generating magnetic flux in a parallel axis, with the first mode and second mode currents 90 degrees out of phase. The surface coil, which supports two resonance current modes for quadrature operation on only one single coil conductor structure, provides a high signal-to-noise ratio (SNR) and a good B.sub.1 homogeneity over the imaging volume. This coil alone may be used either for both transmitting and receiving RF signals or for detecting RF signals as "receive only." This coil is well suited for imaging the human neck, spine and heart.

Number of Drawing Sheets (1):
4

Brief Summary Text (2):
present invention pertains generally to Magnetic Resonance imaging (MRI) apparatus, and more particularly to a quadrature surface coil for use with MRI apparatus.

Brief Summary Text (4):
MRI provides a unique non-invasive imaging method for discriminating the main components of human disease pathology. As a result, MRI is one of the most widely used diagnostic imaging tools in today's hospitals throughout the world. A typical MRI system includes a main magnet to generate a uniform DC magnetic field, three gradient coils to generate linear and orthogonal magnetic field gradients, a transmitting and receiving radio frequency (RF) antenna to generate imaging pulses and receive the resulting RF emissions, and an operator interface and control station. For human imaging the magnet is mainly superconducting in nature and has a cylindrical shape, although at the present time open "C" arm magnet geometries are also used for imaging the human body. For higher strength magnetic fields (0.5 T and higher), the superconducting magnet is used to generate a highly uniform static magnetic field with a clear bore diameter of 90 cm or larger for human patient access.

Brief Summary Text (5):
Gradient coils are electromagnetic coils capable of generating linearly varying and axially directed static magnetic fields along the three spatial directions (x,y,z) of a Cartesian coordinate system. The function of each one of the three orthogonal gradients is to encode the spatial information as a frequency or phase variation. In general, higher gradient strengths of 40-100 mT/m with faster rise times of 40-150 .mu.sec at full strength are required for faster imaging techniques. Standard methods for production of linear magnetic field gradients in MRI systems consist of driving discrete coils with a current source of limited voltage. The discrete coils are wound in a bunched or distributed fashion on an electrically insulating hollow light cylinder coil-form.

Brief Summary Text (6):
A digital radio frequency transmitter transmits radio frequency pulses or pulse packets to a whole body RF coil to deliver RF pulse into the examination region. The RF pulses are used to excite, prepare, saturate, invert, refocus, or manipulate the resultant bulk magnetization due to ensemble average of the magnetic moment of a

specific nuclear spin such as proton in selected portions of the examination region. For whole body applications, the resonance signals are commonly picked up by the same whole body RF coil. For other more regional focused applications, the signals are often picked up by local coils placed at the vicinity of the examination region other than the whole body coil. Alternatively, a receive-only coil can be used to receive resonance signals induced by the body RF coil. For example, in human head imaging, an insertable head coil can be inserted surrounding a human brain at the isocenter of the magnet used for receiving the RF signal. Conventional RF coils for MRI application are the birdcage coil, single loop surface coil and surface array coil.

Brief Summary Text (7):
In MRI, the resultant radio-frequency signals, which are spatially encoded, are picked-up by the receiver RF coil, amplified and then demodulated/digitized by a receiver. A sequence controller controls or schedules the timing sequence of the three orthogonal gradients, RF pulse waveforms, frequency offset, RF phase, data sampling window of the receiver, as well as other events such as triggering to generate a variety of MRI sequences, such as spin echo imaging, gradient echo imaging, fast spin echo imaging, and echo planar imaging. An image reconstruction processor sorts the spatially encoded image data according to the order in which they are received and transforms the data to form the final MR image.

Brief Summary Text (8):
More specifically for RF antenna, a simple conductor loop interrupted with some capacitors of proper values can serve as a local RF antenna that is capable of transmitting RF signal to its vicinity and detecting minute RF signal from its vicinity with a relatively high signal-to-noise ratio (SNR). Such a coil can only transmit and detect one component of the magnetization signal during M imaging, and often is referred as a linear (RF) coil. Since the detectable magnetization for MRI is a two dimensional vector, to improve the SNR and B.sub.1 uniformity, a quadrature version which receives two orthogonal components simultaneously was then introduced for the same geometry configuration.

Brief Summary Text (9):
PRIOR ART QUADRATURE RF COILS

Brief Summary Text (10):
Previously, there were some early quadrature surface RF coil designs described by various people including Hyde, et al., U.S. Pat. No. 4,721,913, issued Jan. 26, 1988, and Mehdizadeh, et al., MRM, 1,256,1988, U.S. Pat. No. 4,918,388, issued Apr. 17, 1990. Both the Hyde and Mehdizadeh designs consist of two separated and isolated coil circuits which simultaneously generate or detect two orthogonal RF polarizations independently. Furthermore, it has been shown that the two circuits can be made mutually decoupled from each other for the respective resonance modes of interests by means of proper geometrical overlapping. The two signals obtained from two orthogonal polarizations can be combined to enhance the SNR in the resulting image using a quadrature combiner. Boskamp et al., in U.S. Pat. No. 5,030,915, issued Jul. 9, 1991, disclose a single circuit quadrature RF coil formed from a pair of parallel mode single loop coils. The Boskamp et al. coil is formed from an annular outer conductor with an inner conductor connected on first and second ends at respective spaced points around the outer conductor.

Brief Summary Text (11):
Another approach to increase both sensitivity and uniformity for a surface coil over a certain coverage was the "half birdcage" design proposed by Ballon et al., JMR, 90, 131-140 (1990). The half birdcage design is also known as a one dimensional ladder resonator. Due to the multiple conductor design of the half birdcage design, it provides better uniformity of field than single loop coils. Although it is an improvement over the single loop design, it is not a quadrature design, lacking the superior SNR of the quadrature RF coils.

Brief Summary Text (14):
The present invention provides a ladder resonator quadrature coil having the superior SNR of quadrature coils and the uniformity of ladder resonator coils. According to one embodiment, the coil of the present invention comprises a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path. According to one aspect of the invention, the coil construction provides a first mode circuit path sensitive to magnetic flux in a first orientation, and a second

mode circuit path sensitive to magnetic flux in a second orientation orthogonal to the first whereby quadrature operation is obtained. According to another aspect, first and second relatively isolated signals representing orthogonal magnetic fields emitted from a body under investigation in an MRI apparatus are obtainable from the respective first and second mode circuit paths. This embodiment, which supports two resonance current modes for quadrature operation on only one single-coil conductor structure, provides a high signal-to-noise ratio (SNR) and a good B.sub.1 homogeneity over the imaging volume. This coil embodiment may be used either for both transmitting and receiving RF signals or for detecting RF signals as "receive only". This embodiment of the invention is also well suited for imaging the human neck, spine and heart.

Brief Summary Text (15):
According to another exemplary embodiment of the invention, the single circuit ladder resonator quadrature coils of the present invention are mounted in combination with a standard birdcage coil to provide additional imaging capability for the neck region of a user having their head positioned in the birdcage coil. According to yet another embodiment, the coils of the present invention are integrated or mounted in combination with other conventional coils to provide supplemental imaging capability.

Drawing Description Text (5):
FIGS. 3 and 4 illustrate simplified schematic circuit diagrams of embodiments of the ladder resonator quadrature surface coils of the present invention with circuit elements for decoupling the coils when used in a receive only mode.

Drawing Description Text (6):
FIG. 5 illustrates a pair of ladder resonator quadrature surface coils of the present invention mounted in combination with a birdcage coil.

Drawing Description Text (7):
FIG. 6 illustrates the coils of FIG. 5 deployed in an MRI apparatus.

Detailed Description Text (3):
The ladder resonator quadrature coil of the present invention combines two separate ladder resonator coils for two polarizations into an integrated coil conductor structure. This new composite coil provides both high SNR and good B.sub.1 homogeneity over the imaging volume. The simplified topological structure of an embodiment of the new ladder resonator quadrature coil design of the present invention is shown schematically in FIG. 1. For simplicity, the decoupling circuits for receive only operation, which are required by the standard MRI system to decouple the coil from the whole body coil by detuning the resonance frequency of the coil during RF transmission, are omitted from the illustration of FIG. 1. As shown in the FIG. 1, the ladder resonator quadrature coil embodiment shown therein can be said to be formed of two ladder resonator coils sharing a common conductor pattern which is symmetrical about a center conductor. In the embodiment of FIG. 1, two parallel conductor strips in the horizontal direction are connected by five (5) or more uniformly spaced vertical oriented conductor strips or legs, also referred to herein as "paths." Both horizontal and vertical conductor strips are interrupted by capacitors. The coil structure is symmetric with respect to the center vertical conductor, or circuit path, to provide an annular first mode circuit path using the two (2) or more spaced conductor legs on each side of the center conductor leg. The circulation of current through this path is shown in FIG. 2A, wherein the dotted center line corresponds to the center leg conductor of coil 1. The first mode circuit path detects or generates magnetic flux of a first polarization from a body under investigation. A second mode circuit path including the center circuit path, the current flows for which are shown in FIG. 2B, detects or generates magnetic flux in another plane of polarization, with the currents in the first and second mode circuit paths 90 degrees out of phase.

Detailed Description Text (4):
Electrically the overall coil embodiment 1 is nothing but a simple one dimensional circuit network for current loops. Such a structure supports many resonance current modes (there are four independent modes), whose mode currents share the same conductor pattern. The current patterns shown in FIG. 2A and 2B for these two useful resonance modes of coil 1 are capable of generating fields either vertical (perpendicular) or horizontal (parallel) to the surface of the coil. Although the frequencies corresponding to the two specific modes are often different, in the coils of the present invention these two frequencies are made identical by

introducing a set of capacitors cross both center vertical conductor (DC) and one of the two end-ring conductors (AC, CB). In order to accomplish the required frequency matching between two polarizations for quadrature operation, the capacitance of the capacitor on the center vertical conductor is varied or set accordingly. In addition, the values for different capacitor components can be predicted for a given structure using a numerical model for the RF coil. At the resonance frequency, two independent RF signals corresponding to the two orthogonal components of the excited magnetization vector can be taken out simultaneously directly from the coil for the purpose of MRI. One RF signal can be taken out from the center vertical conductor strip or leg through a direct electrical coupling cross the capacitor, and the other can be taken out at the center of one of the parallel conductor strips in similar fashion. For example, U.S. Pat. No. 5,030,915 illustrates in more detail how these signals can be taken off of a single circuit quadrature coil. Furthermore, these two useful resonant modes are intrinsically isolated from each other as a result of their corresponding geometric shapes. Other means of tuning the coils are also possible such as also shown in U.S. Pat. No. 5,030,915, and in particular FIGS. 4, 5 and 6 of that patent.

Detailed Description Text (5):
According to one embodiment of the invention, the conductor pattern of the coil 1 can be constructed on an electrically insulated as well as heat resistant former using copper strips etched or applied to the flexible former. The size of the coil can be made larger or smaller depending on the required coverage of a specific application. The face of the coil can be made to be flat, curved or flexible. Using a construction of this type, approximately-21 db isolation (loaded) between the two polarizations is achievable. The two signals corresponding to two orthogonal polarizations were transmitted independently from the coil through two short flexible 50 ohm coaxial cables to corresponding low noise pre-amplification modules. Then, the amplified signals were fed to a multiple channel port on an MRI apparatus for transmission, demodulation, digitization and image reconstruction. With the decoupling circuits as shown for example in FIG. 3 incorporated in a coil of this construction, the resulting coil in a receive only mode was used for imaging studies at 1.5T on a Siemens VisionTM whole body MRI system. While the coil 1 is illustrated in FIG. 1 as having a rectangular shape, it can be any arbitrary shape that preserves the symmetry necessary for tuned operation, such as the general shapes shown in FIGS. 7a, 7b and 7c of U.S. Pat. No. 5,030,915.

Detailed Description Text (6):
With a proper impedance match circuit the two RF signals obtained from the coil structure of FIGS. 1-4 can be pre-amplified and combined using a standard analog quadrature combiner. Alternatively, for achieving the optimal SNR, the MR signals corresponding to two orthogonal polarizations from the coils can each be separately pre-amplified using a low noise RF amplifier and fed into two RF receiver channels, so the two signals are demodulated and digitized independently. The required quadrature signal combination can be accomplished digitally in the image domain using a software combination accompanied with phase correction for compensating the coil reception phase variation. In addition, the shape and geometrical size of the coils made in accordance with the invention can be optimized for a given application.

Detailed Description Text (7):
Furthermore, ladder resonator quadrature coils made according to the present invention can be used completely alone for both transmitting and receiving RF signal as a transmit-receive (T/R) coil for MR imaging and spectroscopy. Using this design, the traditional half-birdcage coil can be modified to be operated in quadrature mode. One of the advantages of the T/R coil design is its relatively simple coil structure, since it does not require decoupling components on the coil; another advantage is the 40% improvement in SNR over conventional birdcage coils.

Detailed Description Text (8):
If a ladder resonator coil made in accordance with the invention is used in a receive-only operational mode, the coil needs to be made invisible to the transmit RF coil during RF power transmit by means of detuning the resonance frequency of the receiver coil away from that of the transmitter coil, which may be a saddle coil or full birdcage coil by way of example. There are various possible approaches suitable for this detuning. Using a few PIN diodes and inductors in proper locations on the coil 1 structure, the resonant frequencies for the two useful modes of coil can be detuned by selectively switching these diodes on or off using an externally applied voltage signal across these diodes. Examples of coils 1' and 1" modified for this

purpose are shown in FIGS. 3 and 4. In FIG. 4, tank circuits activated with the PIN diodes are used to detune the circuit, while in the embodiment of FIG. 3, PIN diodes alone are employed. Of course, many other configurations of circuit elements and position can be used to detune the coil, and the examples given herein are not to be construed as limiting.

Detailed Description Text (9):
For many practical applications, ladder resonator quadrature surface coils 1 (or 1' or 1") can be integrated into or mounted in combination with many other conventional coils available on clinical MRI systems, one of which is a standard birdcage head coil 2 as shown in simplified form in FIG. 5. Such an "integrated" coil set is highly desirable for head-neck imaging applications. This allows the possibility of performing a complete head-neck examination without changing the RF coil in the middle of examination. The different elements can be selectively switched on and off (i.e. detuned) by the means of a DC voltage signal applied to the coils externally. In FIG. 5 coils 1 (or 1' or 1") are shown mounted or integrated with the full birdcage coil 2. Mounting or integration can be accomplished by mechanically or otherwise fixing the location of the coils 1 in relation to the birdcage coil 2 so that, in this case, when a patient's head is positioned in the birdcage coil, their neck is positioned in the sensitive region of the coils 1. By constructing the ladder resonator coils with input and output leads compatible with existing MRI systems, these coils can be made to work with such existing systems, and mounted to work in conjunction with conventional full birdcage coils of various types or manufacture, or in conjunction with other imaging coils commonly used with MRI systems.

Detailed Description Text (10):
FIG. 6 illustrates the use of the coils of the present invention as shown in FIG. 5 when deployed in an MRI apparatus. The MRI apparatus shown in FIG. 6 includes a plurality of superconducting main magnetic field coils 10 to generate a generally uniform static magnetic field along a longitudinal or z-axis of a central bore 12. The superconducting coils are mounted on a dielectric former 14 and received in an annular, helium vessel 16. The helium vessel is filled with liquid helium to maintain the superconducting magnets at their superconducting temperature. A main magnetic field shield coil assembly 18 generates a magnetic field which opposes the fields generated by the main magnets 10 in regions surrounding the superconducting magnets 10. The annular helium reservoir 16 is surrounded by a first cold shield 20 which is maintained at about 20.degree. K. or less. A second cold shield assembly 22 is chilled to about 60.degree.-70 .degree. K. or less. An outer vacuum vessel 24 encases the cold shields to define a vacuum reservoir therearound. Layers of mylar insulation 26 are arranged between the vacuum vessel 24 and the cold shield 22.

Detailed Description Text (11):
A circularly cylindrical, whole body gradient coil assembly 30 is mounted on a circularly cylindrical dielectric former and mounted within the bore 12. A circularly cylindrical, whole body RF coil 32 is mounted on a circularly cylindrical dielectric former and mounted within the bore 12. A circularly cylindrical dielectric cosmetic sleeve 34 shields the RF and gradient coils from view and protects them from damage. The ladder resonator quadrature coil/birdcage head coil combination shown in FIG. 5 is positioned in bore 12 in close proximity to the patient. As illustrated, the ladder resonator quadrature coils 1 (or 1' or 1") and the birdcage coil 36 (in this case a quadrature birdcage coil) each include separate circuits connecting them to the transmitter and receivers. Only one ladder resonator quadrature coil is shown in FIG. 6, but preferably a second is provided on the other side of the patient's neck as shown in FIG. 5, and an additional circuit is provide to transmit and receive RF from the other coil. Alternatively, a switch could be provided to switch between the leads from the two ladder resonator quadrature coils so that only one set of transmit and receive circuits would be needed.

Detailed Description Text (12):
A transmitter 40 is connected with the whole body RF coil 32 for transmitting resonance excitation and manipulation pulses thereto. Preferably, a quadrature divider 42 splits the radio frequency signal into two components and phase shifts one component 90.degree. relative to the other. The two components are applied in quadrature to the whole body RF coil. A gradient control means 44 is connected with the gradient magnetic field coils 30 for providing current pulses thereto for generating magnetic gradient pulses across the examination region. A sequence control means controls the radio frequency transmitter 40 and the gradient control means 44 to generate conventional resonance excitation sequences such as spin echo,

gradient echo, field echo, sequences and the like. Preferably, resonance is excited in two preselected planes or slabs by applying a linear z-gradient field concurrently with a tailored radio frequency excitation pulse for exciting resonance in the two or more preselected slices or slabs. Preferably, one of the slices or slabs intersects the region examined by each of the surface coils 1 (or 1' or 1") and 38.

Detailed Description Text (13):
The radio frequency transmitter 40 and the gradient control 44 under the control of the sequence control 46 elicit simultaneous magnetic resonance responses in planes or slabs through each of the quadrature surface coils 1 (or 1' or 1") and 38. The signals from the two quadrature surface coils are conveyed to a pair of quadrature combiners 50, 52. The quadrature combiners impose a 90.degree. phase shift on one of the detected quadrature components and combine the components. Preamplifiers 54, 56 amplify the signals before they are received by a receiver means 60, such as a pair of digital quadrature receivers 60.sub.1, 60.sub.2, which receive and demodulate the resonance signals. An interface circuit 62 includes analog-to-digital converters 64, 66 for digitizing each received resonance signal to generate a digital data line.

Detailed Description Text (14):
The digital data lines are stored in data memories 70, 72 of a computer means 74. An image reconstruction means such as an inverse two-dimensional Fourier transform means 76 reconstructs sets of data lines from the data memories 70, 72 into electronic digital image representations which are stored in image memories 80, 82. A video processor means 84 converts the digital image representations into the appropriate video format for display on a video monitor 86 or other human-readable display.

Detailed Description Text (15):
Those of skill in the art will recognize that the conductors of the ladder network of the birdcage coil 38 and quadrature coil 1 must be oriented in parallel with the uniform magnetic field, such that use of the coil system with different orientations of the uniform, main field requires reorientation of the coils to meet this requirement.

Detailed Description Text (16):
When the coils of the present invention are used for receiving in the embodiment or arrangement shown in FIGS. 3 and 4, the coil advantageously improves SNR by more closely allowing the correlation of the geometric location of the transmit pulses with the corresponding RF emissions detected by the coils. Furthermore, according to another embodiment, a sensor is provided to detect the pulse in the carotid artery, so that the imaging pulses can be synchronized or triggered on pulse events, and thereby provide repeatable, consistent imaging.

Detailed Description Text (17):
As suggested above, ladder resonator quadrature coils constructed according to the present invention can be of a substantially planar configuration in the manner of a surface coil, or shaped or curved to provide a half-birdcage coil configuration, providing a single half-birdcage coil structure operable in a quadrature mode. In terms of clinical imaging application, coils constructed according to the present invention are well suited for imaging the human neck, spine and heart. For multi-nuclear spectroscopy applications, such coils can be easily made double tuned for simultaneously observing two frequencies.

Detailed Description Text (19):
Further possible modifications to the invention include placing the capacitors on the other end-ring to preserve symmetry. Also, as is well known, inductive coupling can be used for feeding and taking signals out from the coil, or the orthogonal signals can be combined using a standard quadrature hybrid combiner before or after pre-amplification. It is also possible that the two orthogonal modes of the coil can be intentionally tuned to two different frequencies, providing a double tuned coil for multi-frequency or multi-nuclear imaging or spectroscopy.

Detailed Description Text (20):
In addition to the above noted variations in coil design, coils constructed according to the present invention can be made to conform to the geometry of the anatomy using curved (non-straight) vertical and horizontal conductors, or be made to have more open space for the interventional purpose as well as conventional imaging. Furthermore, two or more small coils can be integrated into a ladder

resonator quadrature coil of the present invention for active position and orientation tracking. Moreover, the design concepts of the present invention can be applied to improve many existing MRI surface or wrap-around surface coils for imaging of other parts of human body. Also, although described with respect to imaging the human body, the coils of the present invention can be used to image any animal or other body, and the term "body" or "patient" as used herein should not be interpreted as limited to a human body or patient.

Detailed Description Text (21):
Thus, as described above ladder resonance quadrature coils constructed according to the present invention provide the superior SNR of quadrature coils with the uniformity of field obtainable with ladder resonance coils, in a single circuit design easy to construct and use. As used herein, the term "single-circuit" refers to a circuit connected together with physical electrical components or equivalents of physical components, as opposed to the independent coil designs of prior art quadrature coil systems wherein the coils are formed separately from one another. It should also be noted that many different embodiments of the coil of the present invention are possible, and in particular that the type and placement of the circuit elements necessary for tuning, detuning and quadrature operation can be varied considerably and still achieve the desired operation, as known to those of skill in the art. Although specific embodiments have been illustrated and described herein, it will be appreciated by those of ordinary skill in the art that any arrangement which is calculated to achieve the same purpose may be substituted for the specific embodiment shown. This application is intended to cover any adaptations or variations of the present invention. Therefore, it is intended that this invention be limited only by the claims and the equivalents thereof.

Other Reference Publication (1):
Ballon, D., et al., "A 64 MHz Half-Birdcage Resonator for Clinical Imaging", J. of Magnetic Resonance, 90, 131-140, (1990).

Other Reference Publication (2):
Hu, X., et al., "Reduction of Field of View for Dynamic Imaging", Magnetic Resonance in Medicine, 31, No. 6, 691-694, (1994).

Other Reference Publication (3):
Mehdizadeh, M., "RF Coils for Magnetic Resonance Imaging", RF Design, 29-38, (1991).

Other Reference Publication (4):
Panych, L.P., et al., "A Dynamically Adaptive Imaging Algorithm for Wavelet-Encoded MRI", Magnetic Resonance in Medicine, 32, No. 6, 738-746, (1994).

CLAIMS:

1. A radio frequency (RF) coil construction, comprising a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path.

2. A coil construction according to claim 1 further wherein each ladder resonator coil has two or more conductor paths, in addition to the center conductor path, forming "rungs" of the ladder.

3. A coil construction according to claim 1 wherein a first mode circuit path is sensitive to magnetic flux in a first orientation, and a second mode circuit path is sensitive to magnetic flux in a second orientation orthogonal to the first whereby quadrature operation is obtained.

4. A coil construction according to claim 3 wherein first and second relatively isolated signals representing orthogonal magnetic fields emitted from a body under investigation in an MRI apparatus are obtainable from the respective first and second mode circuit paths.

5. A coil construction according to claim 1 wherein the center conductor path is formed with two or more individual conductors.

6. A coil construction according to claim 1 further including means for detuning the coil temporarily whereby interference with an externally generated RF field is avoided.

9. A coil construction according to claim 1 further wherein the coil has a half-birdcage shape.

10. A radio frequency (RF) coil construction, comprising a pair of first tuned ladder resonator coils each having two or more conductor paths in addition to a shared center conductor path, the ladder resonator coils sharing a common conductor pattern and symmetrical about the center circuit path, said construction providing a first mode circuit path formed to include the conductor paths of each of the ladder resonator coils, the first mode circuit path defining a second coil, the construction including means for tuning the first and second coils to a given RF and means for causing said first and second coils to operate in quadrature and generate first and second relatively isolated signals manifesting said orthogonal magnetic fields in said body.

11. A method of forming a single-circuit quadrature radio frequency (RF) coil, comprising the steps of a) combining a pair of first tuned ladder resonator coils to provide a common conductor pattern and so that they are symmetrical about a center conductor path, and b) adding to the combined coils elements for forming first and second coils each respectively responsive in quadrature mode operation to magnetic fields orthogonal to one another.

12. An integrated radio frequency (RF) coil system comprising a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path fixed in position relative to another coil of a different design, so that the quadrature coil and the coil of a different design can be alternately used when imaging a body under investigation.

13. A coil system according to claim 12 wherein the coil of a different design is a full birdcage coil and the quadrature coil is positioned to have a sensitive region for imaging a neck of a patient whose head is positioned in the full birdcage coil.

14. A method of MRI imaging a patient's neck comprising the step of positioning the patient's head in the birdcage coil of claim 13, and using either the birdcage coil or the quadrature coil for imaging operations.

15. An MRI apparatus comprising a magnet system for generating a steady, uniform magnetic field, a magnet system for generating magnetic gradient fields, a transmit RF coil for generating a RF magnetic alternating field and a surface coil comprising a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path.

16. A method for imaging the head, neck or heart of a patient comprising the step of positioning the coil of claim 1 adjacent thereto and using the quadrature coil in combination with an MRI apparatus to image the same.

L22: Entry 8 of 10 File: USPT Sep 9, 1997

DOCUMENT-IDENTIFIER: US 5664568 A
TITLE: Split-top, neck and head vascular array for magnetic resonance imaging

Abstract Text (1):
A coil assembly (40) includes a first, birdcage type head coil assembly (42) dimensioned to receive a patient's head and a second, neck coil assembly (44) including an anterior coil portion (44a) and a posterior coil portion (44b) dimensioned to receive the patient's neck region. The head and neck coils are partially overlapped. A first cable extends (98a) from the anterior coil portion past the birdcage head coil assembly and a second coaxial cable (98b) extends from the posterior portion past the birdcage head coil assembly. A first decoupling circuit (104a) is disposed in the first coaxial cable beyond a guard ring (106) and a second decoupling circuit (104b) is disposed in the second coaxial cable adjacent the region of overlap between the head and neck coil assemblies. The decoupling circuits are positioned and tuned to prevent radio frequency communication along the coaxial cable sheath between the head and neck coil assemblies. The head and neck coil assemblies are mounted in the mechanical housing which is openable such that an upper half of the guard ring and the birdcage coil and the anterior coil are removable as a unit from the lower half of the guard ring and birdcage coil and the posterior neck coil to facilitate patient access.

Brief Summary Text (2):
The present invention relates to the magnetic resonance arts. It finds particular application in conjunction with split-top insertable radio frequency coils for magnetic resonance imaging of the head and neck and will be described with particular reference thereto. It is to be appreciated, however, that the present invention will also find application in other multiple coil techniques, spectroscopy, phased array coils, imaging for other than medical diagnostic purposes, and the like.

Brief Summary Text (3):
Conventionally, magnetic resonance imaging systems generate a strong, uniform static magnetic field B.sub.0 in a free space or bore of a magnet. This main magnetic field polarizes the nuclear spin system of an object in the bore to be imaged. The polarized object then possess a macroscopic magnetic moment vector pointing in the direction of the main magnetic field. In a superconducting main magnet assembly, annular magnets generate the static magnetic field B.sub.0, along a longitudinal or z-axis of the cylindrical bore.

Brief Summary Text (4):
To generate a magnetic resonance signal, the polarized spin system is excited by applying a radio frequency field B.sub.1, perpendicular to the z-axis. Typically, a radio frequency coil for generating the radio frequency field is mounted inside the bore surrounding the sample or patient. In a transmission mode, the radio frequency coil is pulsed to tip the magnetization of the polarized sample away from the z-axis. As the magnetization precesses around the z-axis back toward alignment, the precessing magnetic moment generates a magnetic resonance signal which is received by the radio frequency coil in a reception mode.

Brief Summary Text (5):
For imaging, a magnetic field gradient coil is pulsed for spatially encoding the magnetization of the sample. Typically, the gradient magnetic field pulses include gradient pulses pointing in the z-direction but changing in magnitude linearly in the x, y, and z-directions, generally denoted G.sub.x, G.sub.y, and G.sub.z, respectively. The gradient magnetic fields are typically produced by a gradient coil

which is located inside the bore of the magnet and outside of the radio frequency coil.

Brief Summary Text (6):
Conventionally, when imaging the torso, a whole body radio frequency coil is used in both transmit and receive modes. By distinction, when imaging the head, neck, shoulders, or an extremity, the whole body coil is often used in the transmission mode to generate the uniform B.sub.1 excitation field and a local coil is used in the receive mode. Placing the local coil close to the imaged region improves the signal-to-noise ratio and the resolution. In some procedures, local coils are used for both transmission and reception. One drawback to local coils it that they tended to be relatively small. The whole body coils are typically used for elongated regions, such as the spine. One technique for adapting surface coils for imaging an elongated region is illustrated in U.S. Pat. No. 4,825,162 of Roemer, in which a series of surface coils are lapped to construct a phased array.

Brief Summary Text (7):
Other radio frequency coil designs include a multi-modal coil known as the "birdcage" coil. See, for example, U.S. Pat. No. 4,692,705 of Hayes. Typically, a birdcage coil has a pair of end rings which are bridged by a plurality of straight segments or legs. In a primary mode, currents in the rings and legs are sinusoidally distributed which results in improved homogeneity along the axis of the coil. Homogeneity along the axis perpendicular to the coil axis can be improved to a certain extent by increasing the number of legs in the coil. Typically, a symmetric birdcage coil has eight-fold symmetry. Such a symmetric birdcage coil with N legs (where N is an even integer) exhibits N/2 mode pairs. The first (N/2)-1 mode pairs are degenerate, while the last mode pair is non-degenerate. The primary mode of such an eight-fold symmetric birdcage coil has two linear modes which are orthogonal to each other. The signals from these two orthogonal or quadrature modes, when combined, provide an increased signal-to-noise on the order of 40%. The simplest driven current pattern or standing waves defined by superpositions of degenerate eigenfunctions. For a low-pass birdcage of N meshes driven at is lowest non-zero eigenfrequency, the current in the n-th mesh is given by sin (2 .pi.n /N+.phi.). The phase angle .phi. determines the polarization plane of the resulting B.sub.1 radio frequency field and can be varied continuously by suitable application of driving voltages. The alignment and isolation of the two linear modes of a birdcage coil are susceptible to sample geometry. That is, the sample dominates the mode alignment and isolation between the two linear modes.

Brief Summary Text (8):
Birdcage coils, like other magnetic field coils, undergo mutual inductive coupling when positioned adjacent each other. As the coils approach each other, the mutual inductive coupling tends to increase until a "critical overlap" is reached. At the critical overlap, the mutual inductance drops to a minimum. As the coils are moved towards a complete coincidence from the critical overlap, the mutual inductive coupling again increases. See, "Optimized Birdcage Resonators For Simultaneous MRI of the Head and Neck", Leussler, SMRM, 12th Annual Meeting, Book of Abstracts, page 1349(1993).

Brief Summary Text (9):
In one multiple coil birdcage design, two birdcage coils have been overlapped to a point of minimum mutual inductance. A symmetric coil is used to image the head and an asymmetric coil was used to image the neck. Capacitive elements are added to provide the necessary phase shifting through the leg sections of unequal length. The coils are mounted in a rigid frame for optimum symmetry and a fixed geometric position. The coils are isolated from each other by the critical overlapping as well as by the addition of neutralization capacitors. See for example, U.S. Pat. No. 4,769,605 issued Sep. 6, 1988 to Timothy R. Fox.

Brief Summary Text (10):
Building an asymmetric coil design is fairly complicated and time-consuming. The phase shift from one section to another in the birdcage coil needs to be maintained for optimum coil performance. The critical overlapping between the two birdcage coils reduces the mutual coupling between the coils to a certain extent. Introducing different coil samples into the two birdcage coils alters the alignment of their linear modes and the mode isolation in either of the coils will change. This change, in turn, affects the symmetry and therefore the mutual coupling between the coils. The greater the mutual coupling between the coils, the larger the noise correlation between the coils and therefore the lower the combined signal-to-noise ratio.

Electrical optimization of such a coil design is very complex. The tuning, matching, and isolation process is iterative and thus time-consuming. More specifically, the two linear modes of each birdcage coil need to be tuned, matched, and aligned to their respective coupling points on the coil and isolated from one another and from the two linear modes of the neck coils. Such a complex and iterative tuning, matching, and isolation process is not readily amenable to mass production.

Brief Summary Text (11):
"Novel Two Channel Volume Array Design For Angiography of the Head and Neck", Reykowski, et al., SMR 2nd Annual Meeting, Book of Abstracts, pp. 216 (1994), discloses a birdcage coil in combination with two volumetric Helmholtz coils arranged such that the B.sub.1 fields of the two Helmholtz coils are diagonal and perpendicular to one another. The two quadrature combined outputs, one from the birdcage coil and one from the Helmholtz coils are interfaced to two channels of the system. By orienting the two Helmholtz coils such that their B.sub.1 fields are orthogonal, coupling is reduced and the noise correlation therebetween held to a minimum. However, when these two volume coils are overlapped with a quadrature head coil, they experience the same difficulties discussed above in conjunction with the multiple birdcage coils. That is, when different sampled geometries are introduced, the isolation between the individual volume Helmholtz coils and the head coil change, causing a change in isolation, resulting in an increased noise correlation between all coils and a lower combined signal-to-noise ratio. Manufacturability of the coil is again complex and time-consuming.

Brief Summary Text (12):
The problem of coil interaction generally exists whenever two or more volume coils of different geometries are used. Electrical optimization of these coil designs is often complicated and iterative, hence time-consuming. Different sample geometries introduced into the designs alter coil-to-coil isolation, resulting in different noise correlations between the coils from one patient to another.

Brief Summary Text (13):
"Head and Neck Vascular Array Coil For MRI", Srinivasan, et al., SMR 2nd Annual Meeting, Book of Abstracts, pp. 1107(1994) and the applicants' co-pending earlier filed related U.S. patent application Ser. No. 08/343,635, filed Nov. 22, 1994 disclose a combination birdcage coil and quadrature volume coil pair. In the described coil design, the coils maintained different current distributions with preferred mode orientations independent of one another. The coil consists of a birdcage coil and a quadrature volume coil pair. The quadrature volume coil pair consists of at least two surface coils of the distributed type, that maintain preferred mode orientations with respect to one another at all times. The birdcage coil maintains an eight-fold symmetry; whereas, the surface coil maintains a two-fold symmetry. After a nominal overlap is achieved between the coils of this design, only one iteration of tuning is required to retune all coils to the magnetic resonance frequency.

Brief Summary Text (14):
The present invention provides a new and improved radio frequency coil which overcomes the above-referenced problems and others.

Brief Summary Text (16):
In accordance with one aspect of the present invention, a magnetic resonance apparatus is provided. A magnet generates a temporally constant, uniform magnetic field through an examination region. At least one radio frequency coil performs at least one of transmitting radio frequency signals into the examination region to induce and manipulate resonance of dipoles therein and receives radio frequency signals from the resonating dipoles. A processor processes the received magnetic resonance signals. The radio frequency coil is characterized by including a first volume coil assembly and a second volume coil assembly. The second volume coil assembly is disposed contiguous to and partially overlapping the first volume coil assembly in a common overlap region. A first electrical circuit is mounted adjacent to and connected with the second volume coil assembly. A first coaxial lead extends from the first electronic circuitry past and contiguous to the first volume coil assembly to a region on the opposite side of the first volume coil assembly from the second volume coil assembly. A first coil-to-coil decoupling circuit is connected with the first coaxial cable for inhibiting the first and second volume coil assemblies from communicating along the first coaxial cable.

Brief Summary Text (19):

In accordance with another more limited aspect of the present invention, the first volume coil assembly includes a birdcage style coil and a guard ring disposed adjacent the birdcage style coil on a side opposite from the first volume coil assembly. The first coil-to-coil decoupling circuit is disposed on an opposite side of the guard ring from the birdcage coil.

Brief Summary Text (21):
In accordance with another aspect of the present invention, a method of magnetic resonance imaging is provided. A temporally constant uniform magnetic field is generated through a head and neck examination region. Magnetic field gradients are applied across the examination region. Radio frequency signals are transmitted into the examination region to induce and manipulate magnetic resonance of dipoles therein. Radio frequency signals are received from the resonating dipoles with a radio frequency coil assembly that has a first volume coil around the head region and a second volume coil around the neck region. The received radio frequency signals are processed into an image representation. The method is further characterized by the magnetic resonance signals from the resonating dipoles in the patient's head and neck regions being received concurrently with the first and second volume coils to generate a volumetric image representation of the head and neck region.

Brief Summary Text (22):
In accordance with a more limited aspect of the present invention, the second volume coil is operated alone to image only the neck region or the first neck coil is operated alone to image only the head region.

Brief Summary Text (23):
In accordance with another aspect of the present invention the first and second coils are tuned individually while separated. The first and second volume coils are overlapped and fixed in their overlapped state. After overlapping, the volume coils are retuned in a single iteration. Thereafter, a patient is received in the volume coils for imaging.

Drawing Description Text (3):
FIG. 1 is a diagrammatic illustration of a magnetic resonance imaging system with an insertable head and neck coil in accordance with the present invention;

Drawing Description Text (4):
FIG. 2 is a perspective view of the head and neck coil of FIG. 1;

Drawing Description Text (5):
FIG. 3 is an electrical diagram and side view of the head coil of FIG. 1 and associated preamplifier and decoupling circuitry;

Drawing Description Text (6):
FIG. 4 is a perspective view of the head coil of FIG. 1 with the top panel removed to illustrate the location of preamplifier, interface, and decoupling circuitry;

Drawing Description Text (7):
FIG. 4A is a detailed view of the anterior neck coil region of FIG. 4;

Drawing Description Text (8):
FIG. 4B is a detailed view of the coupling circuitry region of FIG. 4;

Drawing Description Text (10):
FIG. 6 is a perspective view of the head coil of FIG. 1 viewed from below with the lower casing removed to expose interconnect and decoupling circuitry location;

Detailed Description Text (3):
A whole body gradient coil assembly 30 includes x, y, and z-coils mounted along the bore 12 for generating gradient magnetic fields, G.sub.x, G.sub.y, and G.sub.z. Preferably, the gradient coil assembly is a self-shielded gradient coil that includes primary x, y, and z-coil assemblies 32 potted in a dielectric former and secondary x, y, and z-coil assemblies 34 that are supported on a bore defining cylinder of the vacuum dewar 20. A whole body radio frequency coil 36 is mounted inside the gradient coil assembly 30. A whole body radio frequency shield 38, e.g., copper mesh, is mounted between the whole body RF coil 36 and the gradient coil assembly 30.

Detailed Description Text (4):
An insertable radio frequency coil 40 is removably mounted in the bore in an examination region defined around an isocenter of the magnet 10. In the embodiment of FIG. 1, the insertable radio frequency coil is a head and neck coil for imaging one or both of patient's head and neck.

Detailed Description Text (5):
An operator interface and control station includes a human-readable display, such as a video monitor 52, and an operator input means including a keyboard 54, a mouse 56, a trackball, light pen, or the like. A computer control and reconstruction module 58 includes hardware and software for enabling the operator to select among a plurality of preprogrammed magnetic resonance sequences that are stored in a sequence control memory. A sequence controller 60 controls gradient amplifiers 62 connected with the gradient coil assembly 30 for causing the generation of the G.sub.x, G.sub.y, and G.sub.z gradient magnetic fields at appropriate times during the selected gradient sequence and a digital transmitter 64 which causes a selected one of the whole body and insertable radio frequency coils to generate B.sub.1 radio frequency field pulses at times appropriate to the selected sequence.

Detailed Description Text (6):
Resonance signals received by the coil 40 are demodulated by a digital receiver 66 and stored in a data memory 68. The data from the data memory are reconstructed by a reconstruction or array processor 70 into a volumetric image representation that is stored in an image memory 72. The image is reconstructed from the birdcage and quadrature coil signals when a combined head and neck image is wanted. Alternately, signals from the head and neck coils can be reconstructed separately to make separate images. It is to be appreciated, that for a reconstruction processor and image memory that has a fixed size, e.g., 1024.times.1024.times.1024, that the resolution of the resultant image will be higher when the imaged volume is smaller. A video processor 74 under operator control converts selected portions of the volumetric image representation into slice images, projection images, perspective views, or the like as is conventional in the art for display on the video monitor

Detailed Description Text (7):
With continuing reference to FIG. 1 and further reference to FIGS. 2 and 3, the preferred insertable radio frequency coil 40 includes a birdcage coil 42 and a quadrature coil pair 44 including an upper or anterior coil 44a and a lower or posterior coil 44b. The patient's head is received within the birdcage coil with the anterior coil 44a wrapping around the patient's upper shoulder onto the patient's chest and the posterior coil 44b wrapping around the lower side of the patient's shoulders and along the patient's back. Each of the coils has outputs for two linear modes, preferably orthogonal modes. The birdcage coil and the coils in the quadrature coil pair have capacitors or inductive elements added at appropriate locations such that each operates in a low pass, high pass, band pass, or band stop configuration.

Detailed Description Text (8):
In the preferred split-top embodiment, an upper half of the birdcage coil and the anterior coil are housed in a first or removable housing portion 80. To reduce claustrophobic effects on the patient, the upper housing portion has windows 82 between adjacent legs of the birdcage coil. The upper housing portion is removably received on a lower housing portion 84 which rests on the patient support. The upper and lower housing portions are interconnected by mechanical pins or connectors (not shown). Electrical connectors 86, such as pin, contacts, capacitive couplings, or the like, which may be the same as the mechanical pin or connector assemblies, electrically interconnect end rings 88a, 88b of the birdcage coil 42. Preferably, a mechanical latch 90 holds the first and second portions of the insertable coil assembly together. Adapter tabs or other interconnectors (not shown) are associated with the insertable coil and the patient support to assure accurate alignment of the head coil assembly with the magnetic resonance system. An electrical plug or socket 92 is disposed adjacent a rear end of the insertable coil for interconnection with a matching socket or plug arrangement in the patient support. This enables built-in cable handling assemblies to be provided below the patient support to facilitate operation and use. Although not shown, it is understood that pads are also provided within the coil to immobilize the patient during scanning and to help with patient comfort.

Detailed Description Text (9):
The upper and lower coil portions include formers on which copper foil coils are

supported, e.g., fiber reinforced plastic. The coil formers are fastened to inner portions of the housing for rigidity and for maintaining the coil's shape and structural integrity. Initially, the head coil formers are fixed to the housing. During manufacture and initial calibration, the neck coil formers and the carried neck coil 44 are shifted until a point of minimum mutual inductance with the head coil 42 is reached. At this point, the neck coil formers are fixed to the head coil formers, fixing the relative positions of the neck and head coils. The inner housings are contoured to fit the anatomy under investigation. The outer housings cover the internal parts and electronic assemblies, as well as provide mechanical strength to the coil construction. Between the inner and outer housings, skeletons are provided to add rigidity to the coil structure.

Detailed Description Text (10):
With particular reference to FIGS. 3, 4, 4A and 4B, the head portion of the coil is of a birdcage design. The anterior neck coil 44a is etched on a flexible PC board 94a and mounted on its respective coil former. The coil former and PC board are fastened into the upper housing 80 after the appropriate overlap has been set. The PC board also carries an electronic assembly 96a including matching and decoupling electronics, a preamplifier protection circuit, a preamplifier, and a coaxial cable support assembly. The anterior neck coil is tuned and matched to the magnetic resonance frequency prior to overlapping with the birdcage coil 42. A 50 Ohm coaxial cable 98a passes through an S-shaped or extension region 100a, rides on a support bridge 102a that originates on the anterior neck side and extends along a central plane of the birdcage head coil, and connects to a decoupling circuit 104a. The decoupling circuit is located beyond or rearward of a guard ring 106 of the birdcage coil. The shield of the coaxial cable past the anterior decoupling circuit 104a is connected to the guard ring. As illustrated in FIG. 5, the anterior coaxial cable navigates over the guard ring to the bottom of the coil assembly to an interconnect or output board 108 and the plug or socket

Detailed Description Text (11):
With particular reference to FIGS. 6 and 6A, the posterior neck coil 44b is etched on a circuit board 94b which is fastened to its respective former. The former is adjustably positioned, then fixed to become an integral part of the lower housing portion 84. An electronic circuit 96b is mounted on the posterior coil circuit board 94b. The circuit again includes matching and decoupling electronics, a preamplifier protection circuit, and a preamplifier. A coaxial cable 98b extends from the posterior circuitry 96b through an S-shaped expansion region 100b along a central plane of the birdcage head coil on a bridge 102b to a posterior decoupling circuit which is disposed in the plane of the overlap between the birdcage head coil 42 and the posterior neck coil 44b. The coaxial cable 98b further extends from the decoupling circuit to the interconnect board 108. The shield of the posterior coaxial cable is again connected with the guard ring 106 thus shorting them together.

Detailed Description Text (12):
With reference to FIG. 7, the neck and birdcage coils are each tuned 110, 112 and the orientation of their B.sub.1 fields is adjusted 114, 116. The tuned coils are overlapped 118 a selected amount, e.g., to a point of minimum mutual inductance. The S-shaped extension region 100a, 100b of the cables facilitate ready positioning of circuit boards 94a, 94b relative to the birdcage coil. Once overlapped to the selected point, the coils are fixed together 120. The birdcage coil and the neck coils are retuned 122, 124 in a single iteration to the magnetic resonance frequency. In most cases, the birdcage coil needs most of the retuning and the neck coil little or none. The extent of the retuning depends mainly on the proximity of the coaxial cables 98a and 98b to the birdcage coil 42. In most cases, the neck coil frequencies remain the same. Because only one iteration is needed to retune all coils to the same magnetic resonance frequency after achieving the selected overlap, the number of steps needed for optimization during manufacturing is reduced.

Detailed Description Text (13):
With reference to FIGS. 8A, 8B, and 8C, the anterior and posterior decoupling circuits 104a and 104b each include a housing having a top cover 130a and a bottom cover 130b within which a circular spool 132 is supported. The coaxial cable extends through a first immobilizer or guide 134a, around the spool, and exits by a second immobilizer or guide 134b. Fixed and variable capacitors are soldered across the turn(s) of the coaxial cable to tune the decoupling circuits close to the magnetic resonance frequency. When the top and bottom covers are closed, the immobilizers grip the coaxial cable sufficiently tightly that it is locked against sliding into

or out of the housing. The immobilizers not only reduce any torque from being transmitted to the electronics within the decoupling circuit, but also protect against changes in inductance due to changes in the tightness of the winding around the spool. After the decoupling circuit housing is closed, a sheet of copper foil 136 is wrapped around the housing and the immobilizers. Once the decoupling circuit is positioned in place, the foil is pierced through a small hole 138 to gain access to a trimming capacitor and the decoupling circuit is tuned to the magnetic resonance frequency. After fine tuning, the access opening to the trimming capacitor is foil covered as well. The foil covering functions as a radio frequency shield to isolate the decoupling tank circuit (FIG. 8C) for efficient decoupling circuit operation.

Detailed Description Text (14):
Tuning the decoupling circuits to the magnetic resonance frequency serves two major functions. The primary function is to present a high impedance (Z) for currents flowing in the shields of the coaxial cables during RF transmit. This prevents the formation of closed loops inside the magnet bore. The second function of each decoupling circuit is slightly different. The posterior neck coil has its second mode tuned to the magnetic resonance frequency. The posterior decoupling circuit isolates the currents flowing in the region where the shield of the coaxial cable is exposed to the birdcage coil from the currents flowing in the shield exposed to the posterior neck coil.

Detailed Description Text (15):
For the coaxial cable 98a above the anterior neck coil 44a, the central plane is a plane of symmetry. That is, the central plane is a virtual ground. The anterior neck coil has its first, primary mode tuned to the magnetic resonance frequency. The anterior coaxial cable is transparent to the anterior neck coil. However, when the coaxial cable is guided across the birdcage coil 42, the cable is no longer at a virtual ground plane and currents are induced in its shield. These circulating currents are substantially attenuated by the anterior decoupling circuit 104a. Again, the anterior decoupling circuit is before the guard ring 106 of the birdcage coil. This stops induced RF currents on the shield of the anterior coaxial cable from being communicated to the guard ring through the shorting connection. The guard ring is also broken by shorting capacitors (not shown) to reduce gradient induced eddy currents. The decoupling circuits are shielded to minimize their interaction with the body coil 36 during RF transmit, with the individual coils in the insertable coil 40, and to reduce any irradiation segments of shields of the straight segments of the coaxial cable.

Detailed Description Text (16):
Additional decoupling circuits may also be employed to provide further barriers to the transmission of stray radio frequency signals. If the decoupling circuits are eliminated completely, the coaxial cables would carry currents in their shields. Further, the interaction between the coils in the array would not be minimized. Rather, the two coils would talk to each other through the shield, causing a disadvantageous transfer of noise between the coils. The presence of the decoupling circuits as illustrated maintains the signal-to-noise ratio of the coils in its different operating modes within a few percent. The signal-to-noise ratio and uniformity of the illustrated insertable coil is similar to that of a standard quadrature head coil without neck coils. The signal-to-noise ratio of the posterior neck coil is similar to that of a C-spine element in a cervical-thoracic-lumbar array coil. However, the coverage of the neck coils is greater in the present design.

Detailed Description Text (17):
The coil has three modes--(1) head and neck, (2) head only, and (3) neck only. See TABLE 1 below. In the head and neck mode, the birdcage head coil and the neck coils are operated together to image from the aortic arch to the top of the head. In the head and neck mode, the signal-to-noise ratio is still high as is the B.sub.1 homogeneity. This head and neck imaging mode enables blood flow to be measured and monitored as it flows from the aortic arch beyond the circle of Willis in the brain in a single image. This high signal-to-noise ratio and uniform coverage is important in the imaging of skull-base tumors that are difficult to image with either a head coil alone or a neck coil alone. The neck coils 44 are decoupled during the head only mode and the head coil 42 is decoupled during the neck only mode.

Detailed Description Text (18):
The coils 42, 44 are actively decoupled during body transmit. The coil is interfaced

to the magnetic resonance system via an interface, such as the interface shown in U.S. application Ser. No. 08/286,780, filed Aug. 5, 1994. Individual channel device drivers in the system transmit/receive interface circuit are programmed to provide different sets of voltages in the three operating modes for the insertable radio frequency coil 40, viz, the head only, neck only, and head and neck mode. In the head mode, the neck coils are actively decoupled and only the head coil is resonant at the magnetic resonance frequency. Similarly, in the neck mode, the head coil is actively decoupled and only the neck coils are resonant at the magnetic resonance frequency. In the head and neck mode, all coils in the insertable radio frequency coil 40 are resonant at the larmor frequency and receive magnetic resonance signals.

Detailed Description Text (19):
It is to be appreciated that various linear or quadrature surface coils may be overlapped with the birdcage coil. Such surface coils are preferably shaped in conformity with the surface of the subject adjacent the region of interest.

Detailed Description Text (20):
The electrical circuits 96a, 96b of the preferred embodiment include a preamplifier and output for the anterior neck coil and a preamplifier and output for the posterior neck coil. A birdcage coil output circuit 96c includes two preamplifiers connected to the birdcage coil to provide 90.degree., quadrature outputs. These four preamplified signals in the illustrated embodiment are conveyed to the radio frequency receiver 66 which demodulates the four resonance signals. Alternately, the quadrature resonance signals can be shifted by 90.degree. and combined at the insertable coil rather than after demodulation. As yet another alternative, the signals can be digitized at the surface coil and digital signals sent to the receiver.

Detailed Description Text (21):
In the above-described preferred embodiment, the birdcage coil has eight-fold symmetry and the neck coils two-fold symmetry. However, other symmetries are also contemplated. In the embodiment of FIG. 9, a birdcage coil 140 is again utilized to provide uniform coverage over the brain. A pair of loop-type anterior neck coils 142a, 142b are positioned over the patient. A distributed type posterior neck coil 144 is positioned below the patient's neck. This coil combination is operable in a head-only mode, a neck-only mode, a posterior neck-only mode, an anterior neck and arch-only mode, a left or right interior neck-only mode, a neck and head combined mode, or a combination of the above.

Detailed Description Text (22):
With reference to FIG. 10, a birdcage coil 150 with a pair of quadrature outputs provides uniform head coverage. Two loop type coils 152a, 152b provide anterior neck and arch coverage. Helmholtz and loop type coils 154a, 154b provide uniform posterior neck coverage. The birdcage, the anterior neck coils, and the posterior neck coils may each be interfaced with one or two channels of the receiver to operate singly or in various combinations as discussed above. Other alternate embodiments include birdcage coils that have other than eight-fold symmetry and neck coils with other than two-fold symmetry. The birdcage coil may be circularly cylindrical, elliptically cylindrical, or have other geometries. The neck coil is contoured in such a way as to provide a high signal-to-noise ratio and uniform coverage over its imaging field of view. The coils in the volume quadrature pair can also be of the loop type, Helmholtz type, Figure-8 type, distributed type, or combinations thereof. The signals from the individual coils can be combined prior to or after quadrature combination or prior to or after preamplification. The signals may also be combined digitally post-data acquisition. As another alternative, the individual coils may be tuned to one or more selected magnetic resonance frequencies. In yet another alternate embodiment, the birdcage volume coil is combined with several quadrature pairs in a cascade manner to cover an elongated anatomy under investigation. Decoupling circuits of other designs for inhibiting the flow of radio frequency currents and different numbers of decoupling circuits may also be utilized. The insertable coil need not be in a split mechanical package. For example, the coil may be slid over the patient's head and neck or other portions of the patient's anatomy as may be appropriate to the coil design and application.

Detailed Description Paragraph Table (1):

TABLE 1

Operating Modes	CH1	CH2	CH3	CH4
Head Only	ON	ON	OFF	OFF
Neck Only	OFF	OFF	ON	ON
Head/Neck	ON	ON	ON	ON

Other Reference Publication (1):
"The NMR Phased Array", Roemer, et al., Academic Press, Inc. 1990 Magnetic Resonance in Medicine 16, 192-225 (1990).

Other Reference Publication (2):
"Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck", Leussler, SMRM 1993, p. 1349.

Other Reference Publication (3):
"An Efficient, Highly Homogenous Radiofrequency Coil for Whole-Body NMR Imaging at 1.5 T", Hayes, et al., pp. 622-628.

Other Reference Publication (4):
"Head and Neck Vascular Array Coil For MRI", Srinivasan, et al., Society of Magnetic Resonance, 2nd Annual Meeting, San Francisco, CA (1994) p. 1107.

Other Reference Publication (5):
"Novel Two Channel Volume Array Design for Antiography of the Head and Neck", Reykowski, et al., SMR 2nd Annual Meeting, San Fransisco, CA (1994) p. 216.

Other Reference Publication (6):
"The Asymmetric Birdcage Design: A Quadrature Neck Coil Application", Vij, et al., SMRM 11th Annual Meeting, Berlin, Germany (1992) p. 4010.

Other Reference Publication (8):
"Quadrature-Headcoil and Helmholtz-Type Neckcoil--An Optimized RF Antenna-Pair For Imaging Head, Neck, and C-Spine at 1.0 and 1.5 T", Krause, et al. SMRM 7th Annual Meeting, San Francisco, CA (1988) p. 845.

Other Reference Publication (9):
"Application of High-Order Coils to Surface Coil Imaging of the Lumbar Spine at 1.5 Tesla", Foo, et al., 5th Annual Meeting SMR, Montreal Quebec (1986) pp. 53-54.

CLAIMS:

1. In a magnetic resonance apparatus which includes a magnet for generating a temporally constant, uniform magnetic field through an examination region, a radio frequency coil which performs at least one of (1) transmitting radio frequency signals into the examination region to induce and manipulate resonance of dipoles disposed therein and (2) receiving radio frequency signals from resonating dipoles in the examination region, and a processor for processing the received magnetic resonance signals, the radio frequency coil including:

a first volume coil assembly;

a second volume coil assembly disposed contiguous to and only partially overlapping the first volume coil assembly in a common overlap region;

a first electronic circuit mounted adjacent to and connected with the second volume coil assembly;

a first coaxial cable extending from the first electronic circuit past and contiguous to the first volume coil assembly, to a region on an opposite side of the first volume coil assembly from the second volume coil assembly;

a first coil-to-coil decoupling circuit connected with the first coaxial cable for inhibiting the first and second volume coil assemblies from coupling to each other along the first coaxial cable when receiving the radio frequency signals from the resonating dipoles in the examination region.

2. In the magnetic resonance apparatus as set forth in claim 1, the second coil assembly including a first coil connected with the first coaxial cable and a second coil, the radio frequency coil further including:

a second electronic circuit disposed adjacent to and contiguous with the second coil;

a second coaxial cable extending from the second coil past and contiguous to the

first coil assembly to the opposite side thereof;

a second coil-to-coil decoupling circuit connected with the second coaxial cable for inhibiting the second coil and the first coil assembly from coupling along the second coaxial cable.

3. In the magnetic resonance apparatus as set forth in claim 2, the second coil-to-coil decoupling circuit is disposed in a common plane with the overlap region.

4. In a magnetic resonance apparatus which includes a magnet for generating a temporally constant, uniform magnetic field through an examination region, at least one radio frequency coil which performs at least one of (1) transmitting radio frequency signals into the examination region to induce and manipulate resonance of dipoles disposed therein and (2) receiving radio frequency signals from resonating poles in the examination region, and a processor for processing the received magnetic resonance signals, the radio frequency coil including

a first coil assembly, the first coil assembly including a birdcage style coil and further including a guard ring disposed adjacent the birdcage style coil;

a first electronic circuit mounted adjacent to and connected with the second coil assembly;

a first coaxial cable extending from the first electronic circuit past and contiguous to the first coil assembly, to a region on an opposite side of the first coil assembly;

second coil assembly disposed contiguous to and only partially overlapping the first coil assembly in a common overlap region;

a second electronic circuit disposed adjacent to and contiguous with the second coil;

a second coaxial cable extending from the second coil past and contiguous to the first coil assembly to the opposite side thereof, sheaths of the first and second coaxial cables being electrically connected with the guard ring; and

a first coil-to-coil decoupling circuit connected with the first coaxial cable for inhibiting the first and second coil assemblies from coupling along the first coaxial cable, the first coil-to-coil decoupling circuit being disposed on an opposite side of the guard ring from the birdcage style coil.

5. In the magnetic resonance apparatus as set forth in claim 2, the second coil-to-coil decoupling electronic circuit being disposed substantially in a common plane with the overlap region.

6. In a magnetic resonance apparatus which includes a magnet for generating a temporally constant, uniform magnetic field through an examination region, a radio frequency coil which performs at least one of (1) transmitting radio frequency signals into the examination region to induce and manipulate resonance of dipoles disposed therein and (2) receiving radio frequency signals from resonating dipoles in the examination region, and a processor for processing the received magnetic resonance signals, the radio frequency coil including:

a first coil assembly;

a first electronic circuit mounted adjacent to and connected with the second coil assembly;

a first coaxial cable extending from the first electronic circuit past and contiguous to the first coil assembly;

a second coil assembly disposed contiguous to and partially overlapping the first coil assembly in a common overlap region, including a first coil connected with the first coaxial cable and a second coil;

a second electronic circuit disposed adjacent to and contiguous with the second coil;

a second coaxial cable extending from the second coil past and contiguous to the first coil assembly to an opposite side thereof;

a first coil-to-coil decoupling circuit connected with the first coaxial cable for inhibiting the first and second coil assemblies from coupling along the first coaxial cable;

a second coil-to-coil decoupling circuit connected with the second coaxial cable for inhibiting the second coil and the first coil assembly from coupling along the second coaxial cable;

a lower housing portion which supports the second coil, a lower part of the first coil assembly, the second electronic circuit, and the second decoupling circuit;

an upper housing portion which supports the first coil, an upper part of the first coil assembly, the first electronic circuit, and the first decoupling circuit, the upper and lower housing portions being selectively mechanically couplable to provide a unitary coil construction and decouplable to facilitate patient access; and,

electrical connections for interconnecting the upper and lower parts of the first coil assembly.

7. In a magnetic resonance apparatus which includes a magnet for generating a temporally constant, uniform magnetic field through an examination region, a radio frequency coil which performs at least one of (1) transmitting radio frequency signals into the examination region to induce and manipulate resonance of dipoles disposed therein and (2) receiving radio frequency signals from resonating dipoles in the examination region, and a processor for processing the received magnetic resonance signals, the radio frequency coil including:

a first volume coil assembly;

a second volume coil assembly disposed contiguous to and partially overlapping the first coil assembly in a common overlap region;

an electronic circuit mounted adjacent to and connected with the second coil assembly;

a coaxial cable extending from the first electronic circuit past and contiguous to the first coil assembly, to a region on an opposite side of the first coil assembly from the second coil assembly;

a coil-to-coil decoupling circuit connected with the first coaxial cable for inhibiting the first and second coil assemblies from coupling along the first coaxial cable, the coil-to-coil decoupling circuit including:

a case having a spool-like construction therein, the case having coaxial cable receiving guides for receiving the coaxial cable and locking the coaxial cable against longitudinal movement, the coaxial cable being wrapped around the spool to form an inductance and being interconnected with at least one adjustable tuning capacitor;

an electrically conductive layer for coating and shielding the case.

8. In a magnetic resonance apparatus which includes a magnet for generating a temporally constant, uniform magnetic field through an examination region, a radio frequency coil which performs at least one of (1) transmitting radio frequency signals into the examination region to induce and manipulate resonance of dipoles disposed therein and (2) receiving radio frequency signals from resonating dipoles in the examination region, and a processor for processing the received magnetic resonance signals, the radio frequency coil including:

a first coil assembly, the first coil assembly being at least four-fold symmetric;

a second coil assembly disposed contiguous to and partially overlapping the first coil assembly in a common overlap region, the second coil assembly being at least two-fold symmetric;

an electronic circuit mounted adjacent to and connected with the second coil assembly;

a coaxial lead extending from the electronic circuit past and contiguous to the first coil assembly, to a region on an opposite side of the first coil assembly from the second coil assembly;

a coil-to-coil decoupling circuit connected with the coaxial lead for inhibiting the first and second coil assemblies from coupling along the first coaxial lead.

9. A radio frequency coil for at least receiving magnetic resonance signals, the radio frequency coil comprising:

a first volume coil assembly;

a second volume coil assembly disposed contiguous to and partially overlapping the first volume coil assembly in an overlap region;

a coaxial lead extending from the second volume coil assembly, along the first volume coil assembly, to a region on an opposite side of the first volume coil assembly from the second volume coil assembly;

a coil-to-coil decoupling circuit connected with the coaxial lead for inhibiting the first and second volume coil assemblies from coupling to each other along the coaxial cable while receiving the magnetic resonance signals such that cross-talk between the first and second volume coil assemblies is inhibited.

10. A radio frequency coil for at least receiving magnetic resonance signals, the radio frequency coil comprising:

a first coil assembly, the first coil assembly being at least four-fold symmetric;

a second coil assembly disposed contiguous to and partially overlapping the first coil assembly in an overlap region, the second coil assembly being at least two-fold symmetric;

a coaxial lead extending from the second coil assembly, along the first coil assembly, to a region on an opposite side of the first coil assembly from the second coil assembly;

a coil-to-coil decoupling circuit connected with the coaxial cable for inhibiting the first and second coil assemblies from coupling along the coaxial cable.

11. A radio frequency coil for at least receiving magnetic resonance signals, the radio frequency coil comprising:

a first coil assembly, the first coil assembly including a birdcage coil and a guard ring;

a second coil assembly disposed contiguous to and partially overlapping the birdcage coil in an overlap region, the guard ring disposed adjacent;

a first coaxial cable extending from the second coil assembly, along the birdcage coil, to a region on an opposite side of the birdcage coil from the second coil assembly, a sheath of the first coaxial cable being electrically connected with the guard ring;

a first coil-to-coil decoupling circuit connected with the first coaxial cable for inhibiting the birdcage coil and the second volume coil assembly from coupling along the first coaxial cable.

12. The radio frequency coil as set forth in claim 11, wherein the second coil assembly includes an anterior coil connected with the first coaxial cable and a posterior coil connected with a second coaxial cable, the second coaxial cable extending from the posterior coil along the birdcage coil to the opposite side thereof, a sheath of the second coaxial cable being electrically connected to the guard ring, and further including:

a second coil-to-coil decoupling circuit connected with the second coaxial cable for

inhibiting the posterior coil and the birdcage coil from coupling through the second coaxial cable.

13. The radio frequency coil as set forth in claim 12 wherein the second coil-to-coil decoupling circuit is disposed in a common plane with the overlap region.

14. The radio frequency coil as set forth in claim 13 wherein

the second coil-to-coil decoupling circuit is surrounded by an electrically conductive shield that is electrically connected to the sheath of the second coaxial cable; and,

the first coil-to-coil decoupling circuit is disposed on an opposite side of the guard ring from the birdcage coil and is surrounded by an electrically conductive shield that is electrically connected to the shield of the first coaxial cable.

15. The radio frequency coil as set forth in claim 12, further including:

a lower housing portion which supports the posterior coil, a lower part of the birdcage coil, and a lower part of the guard ring;

an upper housing assembly which supports the anterior coil, an upper part of the birdcage coil, and an upper part of the guard ring, the first and second housing portions being selectively mechanically couplable to provide a unitary construction and decouplable to facilitate patient access;

electrical connectors for providing radio frequency communication between the upper and lower birdcage coil parts and between the upper and lower guard ring parts.

16. A radio frequency coil for receiving magnetic resonance signals, the radio frequency coil comprising:

a first coil assembly;

a second coil assembly disposed contiguous to and partially overlapping the first coil assembly in an overlap region;

a coaxial lead extending from the second coil assembly, along the first coil assembly, to a region on an opposite side of the first coil assembly from the second coil assembly;

a coil-to-coil decoupling circuit connected with the first coaxial cable for inhibiting the first and second coil assemblies from coupling along the coaxial cable, the first decoupling circuit including:

a housing having a spool-like cable guide therein, the housing having first and second coaxial cable receiving shoulders for receiving the coaxial cable and locking the coaxial cable against longitudinal movement, the coaxial cable being looped around the cable guide to form an inductance and being interconnected with at least one tuning capacitor;

an electrically conductive radio frequency shield which shields the housing.

17. In a method of magnetic resonance imaging in which a temporally constant uniform magnetic field is generated through a head and neck examination region, magnetic field gradients are applied across the examination region, radio frequency signals are transmitted into the examination region to induce and manipulate magnetic resonance of dipoles therein, radio frequency signals are received from the resonating dipoles with a radio frequency coil assembly that has a first coil assembly around the head region and a second coil assembly around the neck region, and the received radio frequency signals are processed into an image representation, the improvement comprising:

receiving magnetic resonance signals from resonating dipoles in at least one of the head and neck regions with corresponding first and second volume coil assemblies;

passing signals received by one of the first and second volume coil assemblies through a cable which extends along the other volume coil assembly for processing

into the volumetric image representation;

passing signals received from the other volume coil assembly along a second cable for processing into the volumetric image representation;

preventing the first and second volume coil assemblies from coupling to each other along the first cable;

generating the volumetric image representation of at least one of the head and neck regions from the signals passed along the first and second cables.

18. In the magnetic resonance imaging method as set forth in claim 17, the improvement further comprising:

decoupling the first coil assembly and operating the second coil assembly alone to image only the neck region.

19. In the magnetic resonance imaging method as set forth in claim 17, the improvement further comprising:

decoupling the second coil assembly and operating the first coil assembly alone to image only the head region.

20. In the magnetic resonance imaging method as set forth in claim 17, the improvement further comprising:

operating the first and second assemblies simultaneously to image the neck and head region simultaneously.

21. In the magnetic resonance imaging method as set forth in claim 17, the improvement further comprising:

tuning the first coil assembly and the second coil assembly while the first and second volume coils are separated;

overlapping the first and second coil assemblies and fixing the first and second coil assemblies together in an overlapped state;

retuning the first and second coil assemblies once; and,

positioning a patient's head and neck in the first and second coil assemblies in preparation for inducing magnetic resonance.

22. In a method of magnetic resonance imaging in which a temporally constant uniform magnetic field is generated through a head and neck examination region, magnetic field gradients are applied across the examination region, radio frequency signals are transmitted into the examination region to induce and manipulate magnetic resonance of dipoles therein, radio frequency signals are received from the resonating dipoles with a radio frequency coil assembly that has a first coil assembly around the head region and a second coil assembly around the neck region, at least one coaxial lead extends from the second coil assembly along the first coil assembly and has a radio frequency decoupling circuit of adjustable frequency therein, and the received radio frequency signals are processed into an image representation, the method comprising:

adjusting the frequency of the decoupling circuit to match a received magnetic resonance frequency to prevent the first and second coil assemblies from coupling; and

receiving magnetic resonance signals from resonating dipoles in at least one of the head and neck regions with corresponding first and second coil assemblies to generate a volumetric image representation of at least one of the head and neck regions.

23. A magnetic resonance method comprising:

tuning a birdcage coil which is mounted on a dielectric former to a magnetic resonance frequency;

tuning neck coils that are mounted on dielectric formers to the magnetic resonance frequency;

overlapping the birdcage and neck coils and adjusting the overlap until a preselected degree of mutual coupling is achieved;

fixing the dielectric formers of the birdcage and neck coils in the overlapped position;

retuning the birdcage coil and the neck coils, as necessary, in a single iteration; and,

mounting outer housings around the neck and birdcage coils.

24. The magnetic resonance method as set forth in claim 23 further including:

positioning a head and neck of a patient in the birdcage and neck coils;

exciting resonance of selected dipoles within the patient's head and neck such that the dipoles generate magnetic resonance signals;

receiving the magnetic resonance signals with at least one of the birdcage and neck coils.